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(54) **Title:** MAGNETIC RESONANCE IMAGING DEVICE
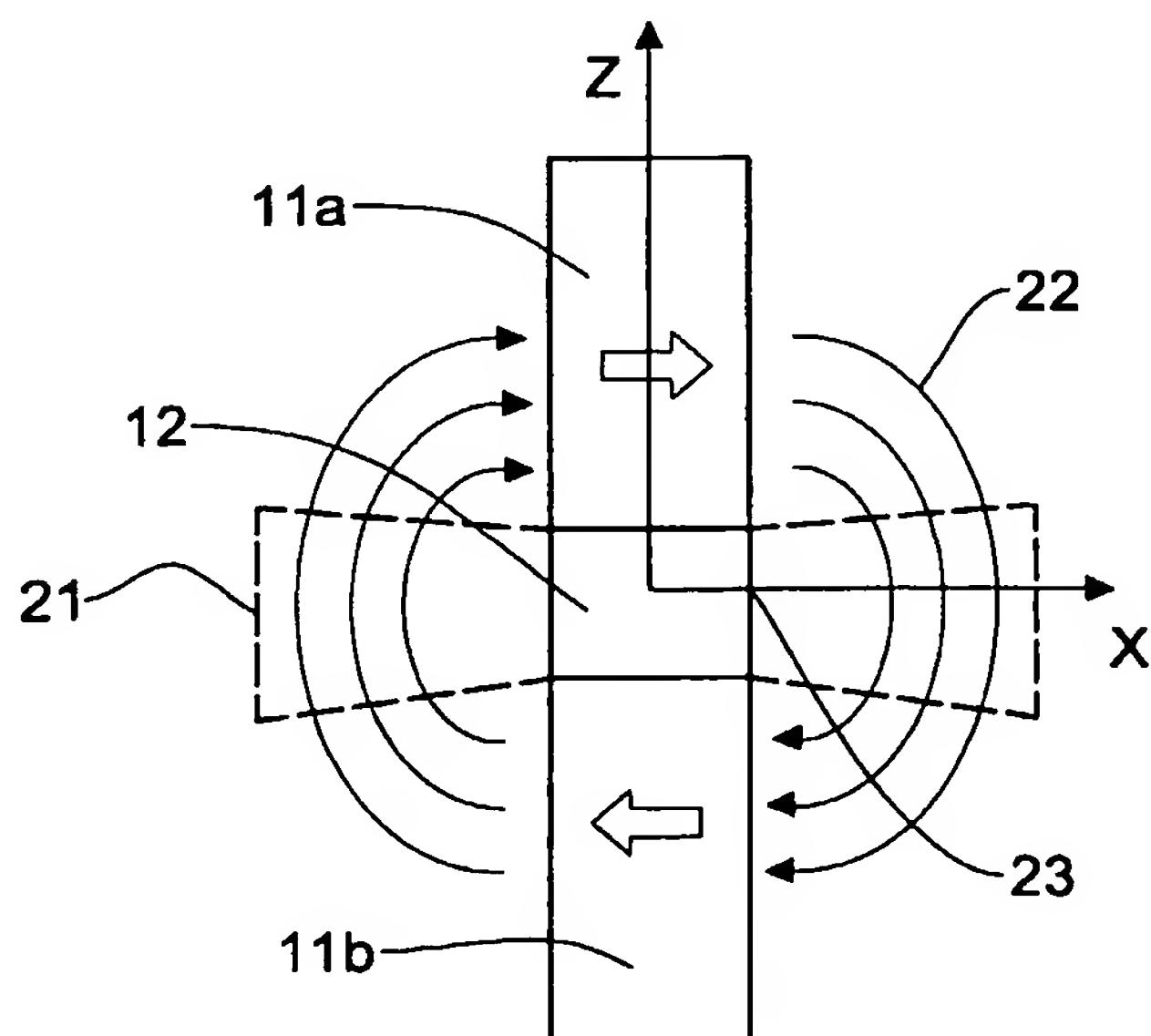
(57) **Abstract:** An imaging probe and method and an MRI system utilizing the same for imaging a region of interest in a medium are presented. The probe comprises first and second magnetic field sources (11a, 11b), and a receiver (12) of NMR signals. The first magnetic field source creates a static magnetic field (22) with a high-gradient magnetic field region in the region of interest outside the probe. The second magnetic field source creates a time-varying magnetic field, which when applied to the high-gradient static magnetic field region, excites nuclei of various resonance frequencies in at least a sub-region of the static magnetic field region to thereby generate the NMR signals. The probe may further comprise a 0-gradient coil (110a, 110b).

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MAGNETIC RESONANCE IMAGING DEVICE

FIELD OF THE INVENTION

This invention is generally in the field of Nuclear Magnetic Resonance (NMR) based techniques, and relates to a device and method for magnetic resonance imaging (MRI). Although not limited thereto, the invention is particularly useful for medical purposes, to acquire images of cavities in a human body, but may also be used in any industrial application.

BACKGROUND OF THE INVENTION

MRI is a known imaging technique, used especially in cases where soft tissues are to be differentiated and identified. Alternative techniques, such as ultrasound or X-ray based techniques, which mostly utilize spatial variations in material density, have inherently limited capabilities in differentiating soft tissues.

NMR is a term used to describe the physical phenomenon in which nuclei, when placed in a static magnetic field, respond to a superimposed alternating magnetic field (e.g., RF radiation). It is known that when the RF magnetic field has a component perpendicular in direction to the static magnetic field, and when this component oscillates at a frequency known as the resonance frequency of the nuclei, then the nuclei can be excited by the RF magnetic field. This excitation is manifested in the temporal behavior of nuclear magnetization following the excitation phase, which in turn can be detected by a reception coil and termed the NMR signal. A key element in the utilization of NMR for imaging purposes is that the resonance frequency, known as the Larmor frequency, has a linear dependence on the intensity of the static magnetic field in which the nuclei reside. By applying a static magnetic field with spatially dependent intensity, it is possible to differentiate signals received from nuclei residing in different magnetic field intensities, and therefore in different spatial locations. The techniques, which utilize NMR phenomena for obtaining spatial distribution images of nuclei and nuclear characteristics, are termed MRI.

In conventional MRI techniques, spatial resolution is achieved by superimposing a stationary or time dependent magnetic field gradient on a static essentially homogeneous magnetic field. By using a series of excitations and signal receptions under various gradient orientations, a complete image of the spatial

5 nuclei distribution can be obtained. Furthermore, it is a unique quality of MRI that the spatial distribution of chemical and physical characteristics of materials, such as biological tissue, can be enhanced and contrasted in many different manners by varying the excitation scheme, known as the MRI sequence, and by using an appropriate signal processing method.

10 In relation to imaging small organs with high resolution, the commercial MRI-based systems suffer from the relatively low signal sensitivity that requires long image acquisition time. Moreover, these systems are expensive and complicated in operation. These drawbacks result from the standard application of the MRI technique, namely, imaging of relatively large volumes, such as the human

15 body. This necessitates producing a highly homogeneous magnetic field and generation of field gradients over the entire imaged volume, leading to extensive equipment size. Additionally, the unavoidable distance between a signal receiving coil and most of the imaging volume significantly reduces imaging sensitivity.

The typical main components of a conventionally used MRI system are as

20 follows: (1) a strong static (DC) magnetic field source that creates a primary static substantially homogeneous magnetic field in the entire volume to be imaged; (2) a signal transmission assembly for transmitting NMR excitation pulses; (3) a signal reception assembly for receiving the so-excited NMR signal; and (4) a static magnetic field gradient source to spatially encode the signals originating from the

25 imaged volume.

The resolution of an image depends on many parameters, some of which are related to the imaging hardware, and some relate to the imaging technique or pulse-sequence used. Practically, however, the resolution is generally governed by two parameters: gradient field strength and signal to noise ratio (SNR) per

30 volume cell (voxel).

There are number of applications in which there is a need for imaging of relatively small volumes, where some of the above-noted shortcomings may be overcome. One such application is geophysical well logging, where the "whole body" MRI approach is obviously impossible. Here, a hole is drilled in the earth's crust and measuring equipment is inserted thereinto for local imaging of the surrounding medium at different depths.

Several methods and apparatuses for extracting NMR data from the bore hole walls have been developed and are disclosed for example in the following publications: US Patents Nos. 4,350,955; 4,629,986; 4,717,877; 4,717,878; 4,717,876; 5,212,447; 5,280,243; and articles "Remote 'Inside Out' NMR", J. Magn. Res., 41, p. 400, 1980; and "Novel NMR Apparatus for investigating an External Sample", Kleinberg et al., J. Magn. Res., 97, p. 466, 1992.

The apparatuses disclosed in the above documents are based on several permanent magnet configurations designed to create relatively homogeneous static magnetic fields in a region external to the apparatus itself. RF coils are typically used in such apparatuses to excite the nuclei in the region of homogeneous static magnetic field and to receive the NMR signal. To create an external region of homogeneous static magnetic field, the magnet configuration have to be carefully designed, so as to reconcile the fact that small deviations in structure may have a disastrous effect on the magnetic field homogeneity. It turns out that such a region of homogeneous magnetic field can be achieved only within a narrow radial distance and small angle variation around a fixed position relative to the magnet configuration, and that the characteristic magnetic field intensities created in this region are generally low. As a result, such apparatuses, although permitting NMR measurements, have only limited use as imaging probes for imaging regions of the bore-hole walls.

With respect to medical MRI-based applications, the potential of using an intra-cavity receiver coil has been investigated, and is disclosed, for example, in the following publications: Kandarpa et al., "J. Vasc. and Interventional Radiology", 4, pp. 419-427, 1993; and US Patent No. 5,699,801. Different designs for

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catheter-based receiver coils are proposed for insertion into body cavities, such as arteries during interventional procedures. These coils, when located in proximity of the region of interest, improve reception sensitivity, thus allowing high-resolution imaging of these regions. Notwithstanding the fact that this approach enables the resolution to be substantially improved, it still suffers from three major drawbacks: (1) the need for the bulky external setup in order to create a static homogeneous magnetic field and to transmit an RF excitation signal; (2) the need to maintain the orientation of the coil axis within certain limits relative to the external magnetic field, in order to insure satisfactory image quality; and (3) excessive heating is generated in the vicinity of the internal coils due to the interaction of gradient fields with the internal electric wires. Because of these three limitations, the concept of intra-cavity receiver coil is only half-way towards designing a fully autonomous intra-cavity imaging probe.

US Patent No. 5,572,132 discloses a concept of combining a static magnetic field source with an RF coil in a self contained intra-cavity medical imaging probe. Here, several permanent magnet configurations are proposed for creating a region of homogeneous static magnetic field external to the imaging probe itself. This approach is somewhat analogous to the concept upon which the bore-hole apparatuses are based. Also disclosed in this patent are several RF and gradient coil configurations that may be integrated in the imaging probe in order to allow autonomous imaging capabilities. The suggested configurations, nevertheless, suffer from the same problems discussed above with respect to the bore-hole apparatuses, namely, a fixed and narrow region of homogeneous static magnetic field to which imaging is limited, and the low magnetic field values characteristic of homogeneous magnetic field configurations.

The main reason why the above-mentioned apparatuses, as well as mostly all NMR or MRI setups, prefer operation with a substantially homogeneous static magnetic field, is that such a setup allows operating in a narrow frequency bandwidth (typically a few Hz). Working in a narrow frequency band allows an easier electronic circuit design for tuning and matching the receiver and transmitter

channels, but, most importantly, it results in low noise or, rather, high signal to noise ratio (SNR) per NMR signal (spin-echo, for example). Owing to the homogeneity of a static magnetic field, the duration of the NMR signal is typically a few milliseconds. The down side of this conventional scheme of working in homogeneous fields is that field homogeneity is hard to achieve over extensive volumes, especially for "inside-out" apparatuses.

SUMMARY OF THE INVENTION

The present invention takes an advantage of a technique described in a co-pending patent application assigned to the assignee of the present application. This technique is based on the operation with substantially non-homogeneous primary (static) magnetic field created by a probe device in a region external to the probe in order to perform MRI of regions outside the probe, which are significantly larger than those obtainable with the prior art devices. The resulting wide frequency bandwidth is more demanding in terms of electronic circuit and receiver/transmitter coil designs.

It should be understood that a static magnetic field created by a magnetic field source of the probe of the present invention is generally non-homogeneous and has a relatively large region of a very high field gradient. The region of the high-gradient magnetic field has dimensions of up to about the dimensions of the probe and is located external to the probe. This approach is different from the conventional one, according to which a homogeneous static magnetic field is created by a magnetic field source, and additional means are used to create a narrow region of a relatively small magnetic field gradient. The term *"high-gradient static magnetic field region"* used herein signifies that a relative change of the static magnetic field intensity, $\Delta B_0/B_0$, is of about 1% and more than the mean value of the static magnetic field intensity B_0 in this region.

The advantageous use of such a high-gradient static magnetic field can be understood from the following. As indicated above, the resonance frequency of a nucleus whose magnetization has been affected by a static magnetic field is directly

dependent upon the static magnetic field intensity. An NMR signal response is generated in the nucleus, if the frequency of an exciting time-varying magnetic field applied to the nucleus corresponds to the resonance frequency of the nucleus. Hence, the use of a high-gradient static magnetic field allows for simultaneously magnetizing a wide range of nucleus locations with different resonance frequencies, and allows using a wider frequency band of a time varying magnetic field (RF field) to efficiently and simultaneously excite nuclear spins and thereby generate NMR signals characterized by a wider NMR frequency bandwidth. This frequency bandwidth is more than 1% (generally, 5%-200%, typically about 10%) of the mean frequency value. It should be understood that the term *"simultaneous excitation"* signifies excitation by a single pulse. This enables to acquire a large number of spin-echo signals (at least one hundred, and typically a few thousands) during one excitation series, the duration of which is roughly limited by a typical spin transverse relaxation time (known as T_2). This stems from the fact that each wideband echo is very short in time, when compared with a narrowband echo. This realization, utilizing special hardware design for high-frequency band operation, enables NMR imaging and measurements over extensive regions of non-homogeneous static magnetic fields.

Thus, comparing the approach of the present invention to the conventional one, the approach utilized in the present invention allows operating with a wider frequency band of simultaneous signal detection, as compared to that utilized in the conventional techniques, which are limited to the frequency bandwidth of 1% of the mean frequency value. A lower SNR per spin-echo (due to a larger frequency bandwidth), as compared to that of the conventional technique, can be compensated for by the possibility to acquire a large number of spin-echo signals (at least one hundred, and typically a few thousands) during one excitation series, the duration of which is roughly limited by a typical spin transverse relaxation time (known as T_2).

The present invention is aimed at designing a novel probe device and an imaging method utilizing this probe, characterized by operation with substantially

non-homogeneous static magnetic field, for use in MRI systems. The probe of the present invention is a fully autonomous intra-cavity MRI probe, comprising all components necessary to allow magnetic resonance measurements and imaging of local surroundings of the probe, obviating the need for external magnetic and electromagnetic field sources.

One of the suitable applications of the invented probe is the imaging of blood vessel walls. The imaging probe in such application is preferably integrated in an intravascular catheter, which is connected to an external imaging console.

The imaging method according to the invention is based on the creation of substantially non-homogeneous (very high gradient) static magnetic field (e.g., by permanent magnets), and the use of wideband reception and transmission channels, as well as specifically designed excitation pulse sequences. The present invention enables to achieve high SNR (similar to that of the conventional MRI techniques) by accumulating a large number of wide-bandwidth short-duration echoes, in a similar overall acquisition time.

For the purpose of better understanding the imaging method according to the present invention, the following should be defined:

The term *"region of interest"* (ROI) signifies a region of the medium to be imaged. For intravascular applications, ROI is typically a segment of the blood vessel wall, extending 360° in circumference, a certain "penetration" distance radially into the vessel wall, and a certain length along the vessel.

The ROI may be formed by a plurality of locally adjacent cross-sectional slices, the so-called "imaging slices", having different positions along the length of the blood vessel (i.e., longitudinal axis of the probe). Each slice may be further divided into angular imaging sectors. Each sector may be imaged by a sequential excitation and signal reception from a set of sub-regions spreading along a radial distance from the probe, each set covering an imaging sector.

There is thus provided according to one aspect of the present invention, a probe device for use in an MRI system for imaging a region of interest in a medium, the probe comprising:

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(a) a first magnetic field source for producing a first, static magnetic field having a profile defining a high-gradient magnetic field region outside the probe such that a relative change of the static magnetic field intensity, $\Delta B_0/B_0$, is of about 1% and more than a mean value of the static magnetic field intensity B_0;

(b) a second magnetic field source operable to create a second, time-varying magnetic field of a wide frequency band, the second magnetic field, when applied to at least a sub-region of said region of the first magnetic field in the medium, simultaneously excites nuclei spins of various resonance frequencies in said sub-region to generate NMR signals characterized by a frequency bandwidth of about 1% and more of a mean frequency value of the NMR signals frequency band; and

(c) a receiver for receiving the NMR signals and generating data indicative thereof.

In order to image the ROI, the probe is placed in the medium such that the static magnetic field region created by the first magnetic field source outside the probe is located within the region of interest.

Preferably, the first magnetic field source comprises a pair of permanent magnet assemblies spaced-apart from each other along the probe axis (Z-axis), wherein the assemblies are magnetized in the opposite directions along an axis (X-axis) perpendicular to the probe axis. The static magnetic field is preferably created in a sector of the imaging slice, defined as the imaging sector. In this sector, the static field is substantially parallel to the Z-axis, and its strength decreases monotonously and substantially as one moves along the X-axis away from the probe edge, and therefore can be characterized by a substantial field gradient along the X-axis. For example, this gradient can be about 30-100T/m within an imaging sector of about 2mm's in the radial dimension.

It should be understood that the non-homogeneity of the static magnetic field along the X-axis will result in short NMR signals (typically spin-echoe signals), namely, significantly shorter than those obtained with the prior art devices

operating within the region of substantially homogeneous primary field. For example, a conventional spin echo may have duration of a few tens of milliseconds, while the spin echoes created by the probe of the present invention may be only a few microseconds wide. This allows the acquisition of a few thousands of spin-echo signals in one excitation series having duration of a typical transverse relaxation time (T_2), which is in the order of milliseconds. Accordingly, the averaging of such a great number of signals enables to increase significantly a signal-to-noise ratio in the final signal to be analyzed. The SNR compensation achieved by averaging, enables the operation with the non-homogeneous static magnetic field, the relative intensity change of which can correspond to a frequency bandwidth of more than 1% of the mean value of a magnetic resonance frequency of nuclei in the static magnetic field region. Practically, the field spread can be up to 200% of mean frequency value and even more. For example, the above-indicated X-gradient static field (30T/m) corresponds to a magnetic resonance frequency range of 3.5MHz to 6MHz, i.e., a bandwidth to center frequency ratio of about 50% along a radial distance of 2mm.

The time-varying magnetic field created by the second magnetic field source (which is formed by at least one RF coil) has sufficiently high intensity and sufficiently wide frequency band, such that it is capable of efficiently and simultaneously exciting nuclear spins to generate NMR signals characterized by NMR frequency bandwidth of more than 1% of their mean frequency value. Accordingly, the receiver has also sufficiently high sensitivity and sufficiently wide frequency band, similar to that of the transmitting coil.

Preferably, at least one RF transmitting coil is a flat solenoid mounted on the probe edge such that the axis of the solenoid coincides with the X-axis.

For obtaining two-dimensional images of an imaging slice in the region of interest, a third magnetic field source may be used for creating the so-called "φ-gradient", which corresponds to an angular coordinate varying magnetic field. Preferably, this third magnetic field source includes a pair of parallel RF solenoidal

coils, which are arranged such that their axis are parallel to the Z-axis, and located in the probe itself and close to the probe edge.

The ϕ-gradient may be created by a magnetic field of approximately linear dependence on the angle in the imaging sector, and can be activated during the imaging sequence. Gradient coil activation can be done in several ways. One alternative is to use double polarity gradient pulses before and after each spin echo, i.e., a positive gradient pulse before echo acquisition, and a similar but opposite (negative) pulse after the echo. In this case, the echo signal is phase-encoded in the angular dimension, and then corrected to the original phase following each echo. By varying the strength or duration of each pair of gradient pulses, while a static radial gradient (along X-axis) exists, a complete scan of k-space can be obtained.

When the imaging of a sector is completed, the probe can be rotated around its axis (Z-axis) in order to image other sectors in the same slice, thereby covering the entire imaging slice. Data obtained from the entire acquisition procedure can be processed in real-time and a two-dimensional image can be reconstructed and displayed on a computer monitor.

In order to create a three-dimensional image of the region of interest, the probe can be moved along the Z-axis or along a channel (if for example the latter is a curve-like). Alternatively, multiple probes can be aligned along the Z-axis, or along a curved-like channel.

According to another aspect of the present invention, there is provided an imaging system comprising an intravascular catheter and a complementary external console associated with electronics for generating transmitted signals and processing received signals, the catheter comprising a fully autonomous probe mounted at the distal end of the catheter for imaging a region of interest surrounding the distal end of the catheter, wherein the probe is constructed as described above.

The electronics mainly includes the following elements: an electronic module for tuning the receiver channel and matching it to the receiver coil, and for pre-amplifying the NMR signal received by said receiver coil, a module for

matching transmission channel to the second magnetic field source, a high-power amplifier for amplifying the transmission pulses to the required high voltage/current, a gradient generator unit for generating gradient field pulses at the required levels, and an external unit for further amplifying the pre-amplified received NMR signal. The system also comprises a connector unit for connecting the intravascular catheter to the external console, both mechanically and electrically. The external imaging console comprises preferably a personal computer (PC) with installed hardware for producing analog transmission signals, for recording digitally the NMR received signals, for processing the signals into images, and for displaying the obtained images on the PC monitor. When the catheter connected to the external imaging console is inserted into the blood vessel, the position of the probe is monitored by available equipment in the catheterization or operating room, such as X-Ray angiography. When the imaging probe is located adjacent to a vessel wall segment, which needs to be imaged, the operator initiates the imaging process via the imaging console controls.

The probe is further connected to an electronic module, which is preferably located in proximity to the probe itself. The electronics module is used to amplify the received signal up to a point where further noise or interference accumulated along the way becomes negligible. The electronics module also comprises basic tuning and matching components of the receiver channel, along with protection components at receiving pre-amplifier inputs. Additional wires connect the probe to following modules of both receiver and transmission channel, which do not have to be located in proximity to the probe. For intravascular applications, it is possible to locate only the probe, pre-amplifying (electronics) and matching capacitor modules in the catheter (which is usually limited in diameter). All other system modules can be located externally to the body and, therefore, with fewer constraints.

According to yet another aspect of the present invention, there is provided an imaging method for creating an image of a region of interest in a medium, the method comprising the steps of:

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(i)   applying first, static magnetic field to the region of interest, wherein said static magnetic field has a high gradient region such that a relative change of the static magnetic field intensity, $\Delta B_0/B_0$, is of about 1% and more of a mean value of the static magnetic field intensity B_0;

(ii)   applying a second, time-varying magnetic field to at least a sub-region of the high-gradient static magnetic field region, wherein said time-varying magnetic field has a wide frequency band such as to simultaneously excite nuclei of various resonance frequencies in said at least sub-region of the static magnetic field region to generate NMR signals characterized by NMR frequency bandwidth of about 1% and more of a mean NMR frequency value;

(iii) receiving the NMR signals and generating data indicative thereof.

BRIEF DESCRIPTION OF THE DRAWINGS

In order to understand the invention and to see how it may be carried out in practice, a preferred embodiment will now be described, by way of non-limiting example only, with reference to the accompanying drawings, in which:

Fig. 1 is a schematic illustration of an MRI system according to the invention;

Fig. 2 more specifically illustrates the internal parts of a catheter used in the system of Fig. 1;

Fig. 3 is a block diagram of a preamplifier used in the system of Fig. 1;

Fig. 4 illustrates a cross-section of the imaging probe shown in Fig. 2;

Fig. 5 illustrates the profile of the Z-component of a primary static field along the X-axis;

Fig. 6 illustrates the contour pattern of a static magnetic field created by the probe in the XY-plane;

Figs. 7 to 9 illustrate the construction and operation of a transmission coil of the probe;

Figs. 10 and 11 illustrate the construction and operation of a receiver coil of the probe;

Fig. 12 illustrates the excitation scheme suitable for acquiring a one-dimensional image of a sector of the region of interest;

Figs. 13 to 15 illustrate the constructional and operational principles of an angular gradient coil assembly suitable to be used in the imaging probe of Fig. 4;

Fig. 16 illustrates the excitation scheme suitable for acquiring a two-dimensional image of an imaging slice of the region of interest;

Fig. 17 illustrates a "raw" pixel in the two-dimensional image obtained with the scheme of Fig. 16; and

Fig. 18 illustrates the excitation scheme suitable for obtaining lateral (angular) separation based on "destructive sharpening" of the region to be imaged.

DETAILED DESCRIPTION OF THE INVENTION

Referring to Fig. 1, there is illustrated an MRI system, generally designated 1, for intravascular imaging utilizing an intravascular catheter 100 that comprises an imaging probe section at the distal end 109 of the catheter, and optionally two inflatable balloons 106. The catheter 100 is brought to a region of interest by "riding" a pre-positioned guide-wire 108. This is a known procedure in interventional cardiology, and therefore need not be described in detail, except to note the following. To fix the relative position of the distal end of the catheter 109 relative to surrounding blood vessel walls 107, the balloons 106 are inflated until sufficient contact is achieved between the balloons 106 and the blood vessel walls 107. The positioning and operation of the imaging probe (which is not specifically shown here) located internally in the imaging probe section of the catheter 100 permits imaging of a ring-shaped slice 21 surrounding the imaging probe in the XY-plane that intersects the vessel walls 107. The slice 21 is a cross-sectional area of the region of interest.

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An image produced at each current location of the center of the probe section **109** with respect to Z-axis (longitudinal axis of the probe) is the image of a two-dimensional cross-sectional area (in the XY-plane) of the vessel walls **107**. The balloons' positions are designed such that the shape distortion at both distal and proximal planes of contact with the wall has a negligible effect on the original wall morphology at the imaging slice **21**.

It should be understood that the use of balloons **106** is one possible solution for stabilizing the relative position of the probe section **109** with respect to vessel walls **107** during the image acquisition (of the order of tens of seconds). The balloons **106** may be of an annular shape that allows continuous circulation of blood therethrough, both during the image acquisition phase and at all other time periods. Lumens can be provided through or adjacent to the probe section **109**, and the imaging probe section **109** may include several inner liquid passageways to permit passage of various liquids, e.g., a contrasting agent, as well as through-flow of blood. All these techniques are known *per se*, and therefore need not be specifically described

Obviously, alternative techniques can be used for fixing the probe section **109** relative to the vessel walls **107**, such as a flexible catheter tip or side-arms (not shown) to push the probe section **109** against the vessel walls **107** and even to move the position of the probe section **109** while touching the vessel walls so that it follows the contour of the walls while scanning the imaging slice. In addition or as an alternative means, it is possible to correct motion effects by either hardware or software means.

The imaging probe itself, as well as other internal catheter parts (not shown here), are connected to an interface unit **102** external to the body. This is implemented by a connector **105** provided at the proximal end of the catheter **100**.

When the mechanical rotation of the imaging probe is required (e.g., when a circumferential image of the entire imaging slice is needed) the interface unit **102** may include a motor drive (not shown) for rotating the imaging probe at the distal end of the catheter around the Z-axis, using an internal torque transferring cable

(not shown). Alternatively, a miniature motor can be located at the distal end of the catheter itself. In such case, the interface unit **102** may include the driving circuitry for the miniature motor. Some of the electronic modules typically used in an imaging system, such as amplification circuits for a receiving channel, matching networks for the transmission/receiving channels, etc., may be incorporated in the interface unit **102**.

The system **1** is further provided with an external imaging console, generally designated **101** connected to the interface unit **102** via a power and control cable **111** and via reception and transmission channel segments **110** and **112**. The external imaging console **101** comprises such main constructional parts as a desktop personal computer (PC) **103** and a high-power transmission amplifier **104**. The PC **103** controls the synchronized transmission pulse generation, signal acquisition, imaging probe rotation and ϕ-gradient activation, all being involved in the imaging process, as will be described further below. In addition, the PC **103** is used as a control station through which an operator interfaces with the system **1**, and for displaying the image.

Fig. 2 more specifically illustrates the internal parts of the imaging catheter **100**. Located at the distal end **109** of the catheter is the imaging probe **10**, which is electrically connected to an electronic module **201**. The imaging probe **10** comprises such main constructional parts as two permanent magnets **11a** and **11b** (constituting together a first magnetic field source) and a coil block **12** that includes a receiver coil and gradient coils (not shown). The magnets **11a** and **11b** are aligned in a spaced-apart relationship along the longitudinal axis (Z-axis) of the probe **10** defining a gap between magnets **11a** and **11b**. In the present example, the permanent magnets **11a** and **11b** have a cylindrical shape. The coil block **12** is located in the gap between the magnets. The construction and operation of the coil block **12** will be described more specifically further below. Further provided in the probe device **10** is a transmission coil **13** (constituting a second magnetic field source), which is attached adjacent to the coil block **12** and has its external surface

shaped as a curved rectangle. The coil **13** is mounted such that its geometrical center coincides with the X-axis, and it therefore faces an imaged sector (not shown), as will be described further below.

5   The electronic module **201** contains electronic components required for tuning/matching of the reception channel and for pre-amplifying NMR signals received from the probe **10**. The pre-amplifying components should preferably be located in proximity to the receiving coil block **12**, in order to minimize effective noise and interference levels. Therefore, the electronics module **201** is typically located 10cm away from the probe **10**.

10   **Fig. 3** illustrates a block diagram of main components of a miniature wide-band electronic preamplifier **EP** constructed and operated according to the invention. The preamplifier is connected directly to the leads of a receiver antenna (coil), to amplify an input signal from the receiver antenna, and covers the bandwidth of 1-10MHz. As shown, the input from the receiver antenna is placed in

15   parallel with a varicap diode D_1, the capacitance of which changes with the DC voltage applied thereto (the so-called "tune voltage"). This enables a quick change in the parallel resonance of the so-obtained L-C circuit, wherein L is the inductance of the antenna and C is the capacitance of the varicap diode D_1. This gives the maximum voltage at frequency $f = \dfrac{1}{2\pi\sqrt{LC}}$. The voltage appearing on this

20   parallel pair is fed directly to the FET amplifier whose output is matched to the next stage amplifier (not shown) already outside the patient's body. As for the other components of the preamplifier **EP**, a capacitor C_1 serves to choke the DC tune voltage in the input of the FET amplifier, and a diode pair D_2 serves to protect the input of the FET amplifier from high voltages induced during the transmission

25   period.

Both the electronic module **201** and the probe **10** are rigid internal parts of the catheter **100**. However, only the probe **10** should be located adjacent to the imaging slice **21** (Fig. 1), which is typically located in a narrow part of the blood vessel. Spacing the electronic module **201** a distance of 10cm from the probe,

allows better maneuverability of the probe **10** into narrow vessel segments, while not substantially sacrificing the overall reception signal-to-noise ratio (SNR).

Turning back to Fig. 2, further provided in the catheter **100** are a capacitor module **202** and a connector **105** to the external interface unit (**102** in Fig. 1). The capacitor module **202** is used for matching the transmission channel network. The physical separation between the electronic module **201** and these components, i.e., the capacitor module **202** and the connector **105**, is effectively unlimited, i.e., about 1-1.5 meters, depending on the catheter requirements.

As better shown in **Fig. 4**, illustrating a cross section of the probe **10** in the XZ-plane (where the transmission coil **13** is not shown for simplicity), the permanent magnets **11a** and **11b** are magnetized along the X-axis, i.e., perpendicularly to the Z-axis, and in opposite directions to each other: the N- and S-poles of the magnet **11a** face the S- and N-poles of the magnet **11b**, respectively. The magnets **11a** and **11b** create a primary static magnetic field B_0 in a region external to the probe **10**, flux lines **22** of the magnetic field B_0 forming closed loops between the magnets. In proximity to the X-axis, the flux lines **22** are directed substantially in the +/-Z direction (i.e., are almost parallel to the Z-axis). The volume (region) surrounding the probe **10** in which the flux lines **22** are directed substantially in the +/-Z direction is defined as the imaging slice **21** (i.e., a sector-like part of the region of interest).

Fig. 5 illustrates the profile of the Z-component of the primary static magnetic field along the X-axis, $B_0^{(z)}(x)$, i.e., along the radial direction with respect to the center of the probe. This profile corresponds to an example of the imaging probe of a 2mm diameter. The distances along the X-axis are measured from the probe edge (**23** in Fig. 4).

As shown, the static field strength $B_0^{(z)}$ decreases significantly with the increase of the distance x from the probe edge **23**. Within the distance of 2mm, the field decreases by about 50%, creating a high static magnetic field gradient $\frac{\partial B_0^{(z)}}{\partial x}$ of

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approximately 30T/m. This high-gradient static magnetic field actually presents a non-homogeneous magnetic field.

It should be noted that, generally, such a 50% decrease of the static magnetic field can be obtained within a distance, which is similar to a transverse dimension of the probe (i.e., the probe's diameter in the present example of the cylindrically-shaped probe 10). For example, a 6mm-diameter probe creates a static magnetic field which drops to 50% of its peak value at around 6mm away from the probe edge. This characteristic of the static magnetic field B_0 is essentially different from that used in the prior art MRI techniques, where typical gradient values are a few Gauss/cm or about 0.05T/m. The consequences of such non-homogeneity of the primary static magnetic field, and the methods by which it can be used as an advantage, will be described further below.

It should be understood that, generally, the exact field profile 24 is a complex function of several parameters, such as magnetic material of which the magnets 11a and 11b are made, their exact shape and size, the size of the inter-magnet gap, etc. The permanent magnets 11a and 11b can be made of a rare-earth magnetic material. In the present example, NdFeB (Neodymium Ferrite Boron) is used, which enables to create a substantially strong static field in the imaging slice 21.

It should be noted that the permanent magnets 11a and 11b and the coil block 12 may have geometrical shapes other than cylinders. A single permanent magnet, as well as more than two such magnets can theoretically be used to construct a primary static field source. The magnetic field source assembly can utilize a soft ferromagnetic material in order to increase field intensity. Generally speaking, any permanent-magnet or electro-magnet based static field source can be used, as long as the profile of the static magnetic field created by this source defines a high-gradient magnetic field as defined above.

Fig. 6 illustrates the static magnetic field pattern in the XY-plane in the vicinity of sector 42 of the imaging slice shown as contour lines 40 delineating zones of equal Z-component of the static magnetic field B_0. Here, the imaging

sector **42** actually presents a portion of the imaging slice at one side of the probe, i.e., within areas along the positive X-axis **41**. The X-axis **41** forms a symmetry axis for both the field pattern and for the imaging slice sector border **42**. Stronger field regions are located near the intersection of the X-axis **41** and the probe edge
5   **23**, while weaker field regions are located further away from the probe edge **23**. The regions of stronger field are concentrated within the paraxial regions of the X-axis (as compared to periphery regions thereof), making the X-axis a preferable imaging axis.

Referring to **Figs. 7 and 8**, the construction and main operational principles
10   of the transmission coil **13** are illustrated. The coil **13** is formed by winding a conducting wire in a flat pattern of windings **50**, or printing a spiral conductor pattern, and then folding it to form a curved rectangle. This transmission coil **13** is then attached to the probe **10**, while its geometrical center coincides with the X-axis **41**. When an electric current is driven through coil leads **51**, and
15   consequently, through the coil windings **50**, a time-varying magnetic field B_1 is created, as shown by flux lines **60** (Fig. 8). This magnetic field B_1 is directed substantially in the X-direction, and its magnitude decreases significantly when moving either away from the X-axis **41**, or away from the probe edge **23** along the X-axis **41**.

20   Nuclear spins residing in the imaging sector **42** are subjected to the static magnetic field B_0 in the Z-direction. The strength of the field affecting the nuclear spins depends on their position in the imaging sector **42**, as delineated by the contours **40** of the static field. When a radio-frequency (RF) pulse is driven into the transmission coil **13**, the transverse oscillating field B_1 is created, which excites the
25   spins that have a resonance frequency corresponding to the pulse frequency. If a substantially short RF pulse is used for excitation, the spins of a relatively wide band of resonance frequencies are excited, i.e., a volume of spins which extends substantially along the X-axis. However, the ability to effectively excite nuclei with resonance frequencies varying substantially from the center frequency of the
30   transmission pulse is limited, and therefore a region, where nuclear spins can be

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effectively excited by a single pulse, can be generally delineated as a "banana"-shaped sub-region **61**.

By varying the center frequency of the transmission pulse, different "banana"-shaped sub-regions of excitation can be selected: a higher center frequency pulse excites a sub-region located closer to the probe edge **23**, while a lower frequency pulse excites a sub-region farther away from the probe edge **23**. This is illustrated in **Fig. 9**, showing three banana-like sub-regions **61a**, **61b** and **61c** sequentially activated by varying the center frequency of the transmission pulse. The pulse width (or shape in general) controls the radial extent of the sub-region: A narrower pulse in the time domain will excite a wider sub-region, as long as the excitation energy of the pulse remains sufficient for the required magnetization tilt angle.

Referring back to Fig. 5, it can be deduced that in order to efficiently excite a relatively wide band of spin resonance frequencies, i.e., about 1/10 of the overall bandwidth corresponding to 0-0.2mm distance from the probe edge **23**, a pulse width of around 1μsec should be used. In order to induce a 90°- or 180°-tilt angle, this pulse has to have power in the order of 100W, and the transmission coil **13** and transmission channel have to be substantially efficient over the entire bandwidth of operation. These requirements should be taken into account in the efficient geometry of the transmission coil **13**, in the characteristics of coil windings **50**, in the design of matching circuits for the transmission channel, and in using a high peak-power amplifier (**104** in Fig. 1) as part of the imaging console (**101** in Fig. 1).

Fig. 10 exemplifies the construction of a receiver coil **70** included in the coil block **12** (the gradient coils are not shown for simplicity). The receiver coil **70** is wound around the central section of the coil block **12**. The receiver coil **70** has a solenoid shape, and is arranged such that the axis of the solenoid is parallel to and coincides with the X-axis **41**. Two receiver coil leads **71** connect the coil block **12** to the electronic module (**201** in Fig. 2). The receiver coil **70** is carefully designed in order to maximize sensitivity to NMR signals to be received. The type and number of coil windings, the coil geometry and position are chosen in order to

maximize its sensitivity over a wide frequency band of operation. Namely, the coil efficiency, termed as B_1/I, which is the magnetic field induced by one ampere current flowing into the coil, is in the order of 0.1T/A at the magnet edge down to 0.005 T/A at 2mm distance (Fig. 11). **Fig. 11** shows a cross section of the receiver

5   coil **70** wound around the coil block **12** in the XY-plane. When an electric current is driven through the coil **70**, a magnetic field is created having flux lines **80** generally directed along the X-axis **41**. From the principle of reciprocity, an oscillating magnetization in any point in space having a component parallel to the virtual flux lines **80** at that same point will be depicted by the receiver coil **70** as

10   induced voltage on the coil leads **71**. Since virtual flux density drops with radial distance from the probe edge **23**, the sensitivity of the receiver coil **70** decreases with this radial distance. The pattern of the receiver coil sensitivity in the XY-plane is shown by contour lines **72** showing the sensitivity decay with the radial distance and with the distance from the X-axis **41**. These contour lines **72** are somewhat

15   similar (although not identical) to the static field contour in the XY-plane (**40** in Fig. 6).

It should be understood that the system components described above are sufficient to perform pseudo-one-dimensional imaging of samples external to and adjacent to the probe **10**. This is implemented by manipulating the catheter **100**

20   such that the probe **10** at its distal part is located adjacent to the imaged sample and such that the X-axis passes along one dimension of the sample.

An example of the imaging method will now be described with reference to Figs. 8 and 12. This method is based on a sequence of the Carr-Purcell-Meiboom-Gill (CPMG) sequence type. The terminology and graphical

25   method used to describe this sequence is known *per se*, and therefore need not be specifically described. In **Fig. 12**, a graph **161** corresponds to the time-base envelope of the RF pulses generated by the transmission coil **13** to excite the nuclei, and a graph **162** corresponds to a qualitative sketch of the expected magnetic resonance signals (echo signals). A series **163** of the wide-band (or

30   "hard") RF pulses is preferably used to simultaneously excite nuclei in one

sub-region **61** of the imaging slice sector **42** (Fig. 8), so as to obtain a series of spin-echoes **167**. The position of the banana-like sub-region **61** along the X-axis is selected by setting the RF frequency of the pulses **165** and **166** to a value representing the typical resonance frequency F_1 of nuclear spins in that sub-region.

5   One possible example for obtaining the spin-echoes **167** is the transmission of a 90°-RF pulse **165** followed by a series of 180°-RF pulses **166**, using the time delays τ and 2τ, respectively, shown in graph **161**. It should, however, be noted that numerous other excitation schemes are applicable for obtaining a series of spin-echoes **167**, and can therefore be used for the purposes of the present

10   invention. Sequences based on magnetization tilt angles different than 90°and 180°, on stimulated echoes, and on variations in pulse timing and phase are few examples of many different sequences well known in the art, which can be used in accordance with the present invention.

The spin-spreading, which is rephased at each echo center, stems

15   predominantly from the strong X-gradient in the static magnetic field, which exists permanently in the imaging sector **42**. In the example of Fig. 5, this gradient can be estimated at around 30T/m corresponding to a magnetic resonance frequency range of 3.5MHz-6MHz for nuclei in the imaging sector **42** of a 2mm radial dimension. Even for a single sub-region of sufficient radial extent, the simultaneous resonance

20   frequency bandwidth can be of the order of 0.5MHz. It is thus evident that the method according to the invention enables to simultaneously operate within a substantially wide frequency bandwidth with respect to the mean frequency value of the magnetic resonance signals. In the present example, the overall frequency bandwidth is approximately 50% of the center frequency, while the simultaneously

25   excited frequency bandwidth (corresponding to a single sub-region) may reach 5-10% of the center frequency. In general, the method and design of the different components disclosed as part of this invention, enables to successfully operate in the non-homogeneous magnetic field that corresponds to the non-conventional frequency bandwidth of more than 1%, and typically 10%, of the mean value.

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It will be understood that one-dimensional (1-D) spin-density profiles of the imaging sector **42** can readily be obtained by sampling the echo signal (**167a** in Fig. 12, for example) using typical τ values and acquisition window sizes of several microseconds, and transforming it using the conventional Fourier transform technique. The fact that the static magnetic field profile (**24** in Fig. 5) is not perfectly linear can be readily compensated for during post-processing.

The relatively short τ values makes it possible to acquire a few thousands of spin-echoes **167** in one excitation series **163**, having duration of a typical transverse relaxation time (T_2). The multiplicity of spin-echoes **167** acquired can be averaged prior to any transformation in order to improve the signal-to-noise ratio (SNR) and, as a result, the quality of the final image.

When signal acquisition from one sub-region (**61** in Fig. 9) is completed, the next sub-region can be acquired by repeating the above pulse burst **167** using a new pulse burst **164** with a different center frequency (F_2) in the transmitted pulses. By properly designing the order of excited sub-regions, and by shaping the RF pulses, the excitation of one sub-region can be such as to produce little or ineffective excitation of neighboring sub-regions. This allows for sequential excitation of sub-regions without the requirement for the period of burst repetition T_F to be greater than T_1, or, in other words, allows for the acquisition of a complete set of sub-regions covering the entire imaging sector **42** in one T_1 time frame.

Several methods are available for obtaining two-dimensional (2-D) images of the imaging slice sector **42**. For example, an integrated ϕ-gradient coil can be utilized. Such a coil may be of a conventional type, for example, as disclosed in the above-indicated US Patent No. 5,572,132.

In the present example illustrated in **Figs. 13 and 14**, a ϕ-gradient coil is designed to control the gradient profile in the imaging sector **42**. This gradient coil is formed by two solenoids **110a** and **110b** wound around two rods of ferrite material **121a** and **121b**. The directions of coil windings are opposed, i.e., one coil **110a** is wound in a right hand helix, while the other coil **110b** is wound in a left hand helix. Both coils **110a** and **110b** are located in the coil block **12**, together with

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the receiver coil (not shown here for simplicity). The described gradient coil and winding geometry is anti-symmetric with respect to the X-axis, as can be seen in the cross-sectional XY-plane view of Fig. 14. The two ferrite rods **121a** and **121b** and wound solenoids **110a** and **110b** can be observed in the coil block **12** proximal to the probe edge.

It should be noted that the provision of ferrite rods **121a** and **121b** is optional, and is used to improve the efficiency of the gradient field by concentrating the magnetic flux lines in the solenoids, thereby improving the field intensity per unit current driven in the coils **110a** and **110b**.

The coil leads (not shown) are electrically connected (in series or in parallel) to the gradient generation module (**113** in Fig. 1), which is part of the imaging console (**101** in Fig. 1). When an electric current is driven through the coils **110a** and **110b**, a magnetic field is created in the imaging sector **42**, which, proximal to the XY-plane, is directed substantially in the +/-Z direction. The strength of this magnetic field is an anti-symmetric function of the ϕ-coordinate, as delineated by the contours **120**.

This way, a ϕ-gradient field, $\partial B^{(z)}_{grad}/\partial\phi$, is created in the imaging sector **42**. A profile of the created gradient field along the ϕ-coordinate is illustrated in Fig. 15, showing the Z-component of the gradient field B_{grad} as the function of the ϕ-coordinate. It can be observed that over an angular sector of around 60° (+/-30°), which corresponds to the extent of the imaging sector **42**, B_{grad} is a monotonic and close to linear function of the ϕ-position.

It should be understood that a great variety of two-dimensional imaging sequences can be used in accordance with the proposed embodiment of gradient coils. One possible example of such sequence is illustrated in Fig. **16**. The sequence is generally based on the above-described one-dimensional sequence of Fig. 12, namely, the transmission RF pulse envelopes **141**, received signal **142** and gradient pulses **150**. In the sequence outlined in Fig. 16, RF pulse bursts **143** and **144** are transmitted using the same center frequency. The center frequency can be changed in between groups of pulse bursts, as will be described below. Gradient

pulses are transmitted in pairs in between every two 180°-pulses **146**, before and following each spin-echo signal acquisition segment **147**. The first (positive) gradient pulse **151** follows the first 180°-pulse **146a,** and phase-encodes the spins in the ϕ-direction. The phase-encoded echo **147a** is then acquired. The second

5   (negative) gradient pulse **152** reverses the phase shifts generated by the first gradient pulse **151**, prior to the next 180°-pulse **146b**. The gradient pulse levels **148** and **148a** remain constant throughout each pulse burst **143**, and the echoes acquired during this burst can be averaged for improved SNR.

During the next pulse burst **144,** the gradient pulse levels **149** and **149a** are

10   changed and the process is repeated. By repeating the process with N different (and preferably linearly varying) gradient levels, a scan of k-space in two dimensions, which are roughly the radial dimension (X-axis) and the angle dimension (ϕ), can be achieved. By applying the two-dimensional Fourier transform with appropriate weighting and calibrations, a two dimensional image of the sample in a single sub

15   region (**61** in Fig. 9) of the imaging sector can be obtained.

Referring now to **Fig. 17,** pixels **130** in this two-dimensional image in the XY-plane are schematically illustrated. The contours **40** of the static field B_0 are shown together with the contours **120** of the ϕ-gradient field B_{grad}. The pixel **130** is defined by two consecutive contours of the B_0-field in the first (radial) dimension

20   and by two consecutive contours of B_{grad}-field in the second (ϕ) dimension. This pixel **130** is actually the so-called "raw" pixel, since it will eventually need to be translated by interpolation and calibration into more conventional polar coordinates R and ϕ along the axes **131** and **132**, respectively.

Alternative methods for obtaining lateral (ϕ) separation can be based on

25   "destructive sharpening" of the sub-region **61** in the ϕ-dimension. This effect can be achieved by using the ϕ-gradient coil **110** design, which produces a substantial magnetic field in regions of large (absolute) ϕ-values (Fig. 15). These substantial fields can be used for creating destructive dephasing of spin-echo components coming from nuclear spins in regions of large (absolute) ϕ-values. Spins located at

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$|\phi|>\Delta\phi_0$ are intentionally dephased, in contrast to spins located at $|\phi|<\Delta\phi_0$, wherein $2\Delta\phi_0$ is the required angular resolution.

One possible example for a sequence capable of producing the above effect is shown in **Fig. 18**. This sequence **180** is generally similar to the sequence shown in Fig. 16, but utilizes a different transmission scheme of gradient pulses **191** and **192**: A single gradient pulse is transmitted immediately prior to each of the 180°-pulses **186a, 186b** and **186c**, and these gradient pulses have alternating polarities, i.e., the first gradient pulse is transmitted with a positive (or negative) sign, the second gradient pulse is transmitted with a negative (or positive) sign, etc.

It can be shown that the magnetization phase in the XY-plane following such a sequence of gradient pulses is accumulated to a value, which substantially grows with the absolute gradient field value at the location of the magnetization vector. Angular regions characterized by substantial gradient field values will accumulate a phase, which rapidly varies over angular position, effectively destroying any contribution to the signal from that region. On the other hand, nuclear spins positioned in the vicinity of the X-axis **41** accumulate a very small phase, and therefore contribute substantially to the echo signal. By this, the reception sub-region is "sharpened" in the angular dimension.

An alternative technique for obtaining lateral (ϕ) separation, which may obviate the need for a ϕ-gradient coil, utilizes processing of phase variations in the received signals acquired from overlapping imaging sectors (not shown), as the imaging probe **10** is rotated around the Z-axis.

In order to image the entire imaging sector **42**, any of the above two-dimensional imaging techniques can be repeated for the different sub-regions (**61a, 61b** and **61c** in Fig. 11) covering the imaging sector. Selection of the sub-region is done using selection of transmission pulse frequency and preferably shaping the pulse to better define the sub-region limits.

In order to obtain a complete 360° two-dimensional image of the sample surrounding the probe **10**, the probe itself is preferably rotated around its axis (the Z-axis).

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It should be understood that a proper calibration of the received echo signals is required as part of the signal processing algorithm. This is due to the fact that the signal corresponding to each frequency component of the echo is a complex function involving the relative volume of spins with corresponding resonance

5   frequency, the transmission effectiveness and reception sensitivity typical to that volume, etc. These calibrations can be readily designed and carried out such that the obtained profile correctly represents the distribution of spin density (or any other NMR parameter) along the radial vector (**131** in Fig. 17).

Turning back to Fig. 1, the system according to the invention operates in the

10   following manner. The operator initiates the image acquisition process by running the imaging sequence and collecting received signals, while the imaging probe is rotated. The first magnetic field source (permanent magnets **11a** and **11b**) creates a static non-homogeneous (very high gradient) magnetic field having a sufficient strength (of about 1/10 Tesla) in the region of interest outside the probe. This

15   magnetic field region can be visualized as a sector of a plane perpendicular to the catheter axis (imaging slice). The second magnetic field source (RF coil) is used to excite the nuclei located in a sub-region of said sector of the imaging slice, while the receiver is used to receive the NMR signal (typically an echo-signal) produced by the nuclei.

20   In order to produce a two-dimensional image of the imaging slice, two magnetic field gradients are used: (1) generally radial in direction and inherent in the static magnetic field profile, which generally drops in strength when moving away from the probe edge; and (2) time-dependent and directed generally perpendicularly to the radial gradient (i.e. the angle direction).

25   A sequence of excitation pulses (transmission pulses), preferably a multiple spin echo sequence, such as the well known CPMG, is transmitted by the second magnetic field source (RF coil). This sequence has the advantage that a large number of spin-echoes can be recorded during a time frame of signal coherence, even at substantial gradients of the static and RF fields. In the specific example of

30   the invention, excitation of nuclei within the sector-shaped region is done by

dividing this region into sub-regions, such that each sub-region includes nuclei corresponding to a narrower resonance frequency band, as compared to that of the entire sector-shaped region. Different sub-regions of the sector-shaped region are located at different radial distances from the probe edge. In order to excite a

5   specific sub-region, the carrier frequency of each transmission pulse burst generated by the RF magnetic field source is determined according to the specific resonance frequency of the nuclei residing in that sub-region. By stepping the carrier frequency from one pulse burst to the next, the entire sector-shaped region at the current location of the probe is scanned. Using extremely short pulses of

10 sufficient power allows each sub-region to be substantially wide, meaning that a single pulse can excite nuclei in a substantially wide radial distance. Therefore, an image of the entire sector-shaped region can be acquired by a few pulse bursts, and by rotating the probe the entire imaging slice can be scanned and imaged.

Transmission pulses are generated at low level voltages by specific

15 hardware located in the PC **103** of the imaging console **101**. Then, the pulses are amplified by the high-power amplifier **104**, and driven via transmission channel segment **112** and the interface unit **102** into the transmission channel in the catheter **100** itself. Gradient pulses are generated by the gradient generator module **113** as required by the sequence running on the PC **103**, and are driven into the gradient

20 channel in the catheter **100**. NMR signals received by the imaging probe in the catheter are amplified by the interface unit **102** and transferred via the receiver channel segment **110** into data acquisition hardware in the PC **103**. Acquired data is stored and processed in real time in the PC **103** memory.

At the end of the acquisition phase, the data is processed to produce a

25 high-resolution image of a cross section of the vessel walls **107** contained in the imaging slice **21**, which can be displayed on the PC **103** monitor. The NMR signal received in each spin echo can be readily transformed into a one-dimensional density profile (neglecting NMR relaxation and diffusion effects) of the nuclei in the excited sub-region, along a radial vector. This is because each echo is acquired

30 under a static (inherent) radial field gradient. For obtaining two-dimensional

images, the angular gradient coils, which create a magnetic field of approximately linear dependence on the angle in the imaging sector, are activated in the imaging sequence. This is implemented either by using double-polarity gradient pulses before and after each spin echo (to thereby obtain an echo signal phase-encoded in the angular dimension), or by varying the strength or duration of each pair of gradient pulses, while a static radial gradient exists (to thereby obtain a complete scan of the k-space). When imaging of the sector-shaped region is completed, the probe can be rotated around its (and the catheter's) axis in order to image other sectors, therefore covering the entire imaging slice. Multiple cross-section images can readily be collected by moving the imaging probe section **109** along the Z-axis. It should be noted, although not specifically shown, that the same can be achieved by using an array or "stack" of imaging probes aligned along the Z-axis, or mounted along a flexible pin-like support, as the case may be. The series of cross-section images obtained by either of the above techniques can be further processed to reconstruct 3-D images of the vessel.

The external modules operate in the following way: The pulse sequence is generated in the PC using digital to analog hardware. The analog transmission pulses are amplified by the high power amplifier and driven into the transmission coil inside the catheter via the interface unit and transmission tuning capacitor. The NMR signals received by the receiver coil are pre-amplified by the pre-amplifying unit in the catheter and then by additional amplification stages outside the catheter. The analog received signal is digitized in the PC and acquired into the memory. It is then processed in order to give a full two- (or three-) dimensional image, which is displayed on the PC monitor. The pulse sequence generated by the PC also controls the gradient field pulse generation using the gradient generator module.

In accordance with the present invention, the MRI probe enables imaging of a blood vessel from within the vessel lumen, or the walls of any cavity in the human body, or in any other industrial application. Adaptation of the invention for a specific application can be done by variations in most the imaging probe

characteristics. Additionally, the imaging method can be varied in accordance with the image requirements of the specific application.

The advantages of the present invention are thus self-evident. The present invention utilizes a very high gradient static magnetic field (non-homogeneous magnetic field) in an extensive region of interest external to the probe. This static magnetic field may be produced either externally or internally to the field sources. The use of such non-homogeneous static magnetic field enables simultaneous excitation of nuclei located at different locations in the static field region using a wide frequency band of time varying magnetic field. In addition, the imaged region (sector) of the region of interest (ROI) is substantially extensive, relative to the dimensions of the field sources in general, or, when the sources are contained in an imaging probe, relative to the dimensions of the imaging probe.

The hardware characteristics and imaging method used in the present invention are very unique and non-conventional in the context of ordinary MRI. Indeed, designs of both the transmission and the receiver channels are such that these channels are operable in a simultaneous wide band of frequencies. The extremely sensitive receiver channel includes an extremely low noise pre-amplifier and is matched to a sensitivity-optimized coil over a wide frequency band. The requirement for high sensitivity stems from the requirement to overcome the high noise levels characterizing the wide frequency band. The imaging sequences utilize the short echo duration, again – following the wide frequency band. The short echo duration allows averaging of thousands of echoes in a single coherence duration (T_2) of a few tens of milliseconds. This allows averaging and, therefore, a significant improvement in the final SNR.

It is important to note that according to the invention, the static magnetic field source itself serves as a means for creating a magnetic field gradient to image the region of interest along one dimension thereof. A gradient creation means are typically utilized in an MRI system, and always as a separate element, in addition to static and time-varying RF magnetic field sources.

Hence, the non-homogeneity of the static magnetic field used in the present invention allows for advantageous use of several hundreds and more echo averaging for imaging purposes, and for improving the detection of signals originating from nuclei in materials characterized by relatively short T_2 values. The provision of the high sensitivity reception coil and channel enables to significantly improve the image resolution. Combination of the static field gradient inherent in the static non-homogeneous field, together with the integrated gradient coils and rotational motion of the probe allow separation in both radial and angular dimensions and the creation of two-dimensional images in the plane perpendicular to the probe axis.

It should be understood that the present invention as described above is not limited specifically to producing two-dimensional images of imaging slices and then "piling" them up for three-dimensional images of the ROI, but rather allows manipulation of the received NMR data in many different ways. Data received from some sub-regions can, for example, be used for tissue characterization and display of parameters such as chemical compositions, while data from other sub-regions may be processed to create a tissue contrast image. The resolution may be different in different sub-regions, meaning: "raw" pixels can have different sizes and shapes and the data received from the different pixels can be manipulated and interpolated to create position-dependent effective resolution, e.g., better resolution at shorter radial distances.

Furthermore, a scan of the region of interest can be carried out along the Z-axis without rotating the probe, therefore creating an X-Z plane (longitudinal) image of the vessel, or a Z-scan with display of specific tissue parameters as a function of Z-position. The detection of specific atherosclerotic plaque types can be done by scanning vessel segments and measuring specific parameters indicative thereof.

CLAIMS:

1.   An imaging probe for use in an MRI system for imaging a region of interest, the probe comprising:

(a) a first magnetic field source for producing a first, static magnetic field having a profile defining a high gradient magnetic field region outside the probe such that a relative change of the static magnetic field intensity, $\Delta B_0/B_0$, is of about 1% and more than a mean value of the static magnetic field intensity B_0;

(b)   a second magnetic field source operable to create a second, time-varying magnetic field of a wide frequency band, the second magnetic field, when applied to at least a sub-region of said static magnetic field region in the medium, simultaneously excites nuclei spins of various resonance frequencies in said sub-region of the medium to generate NMR signals characterized by a frequency bandwidth of 1% and more of a mean frequency value of the NMR signals frequency band; and

(c) a receiver for receiving the NMR signals and generating data indicative thereof.

2.   The probe according to Claim 1, wherein said first magnetic field source comprises first and second permanent magnet assemblies aligned in a spaced-apart relationship along a longitudinal axis of the probe defining a gap between the permanent magnet assemblies, the first and second permanent magnet assemblies are magnetized in two opposite directions along an axis perpendicular to said longitudinal axis of the probe, said static magnetic field being thereby parallel to said longitudinal axis and being characterized by the magnetic field gradient along the axis of magnetization of the magnets.

3.   The probe according to Claim 2, wherein said high-gradient static magnetic field region presents an imaging slice in the region of interest, the slice surrounding

the first magnetic field source in a plane perpendicular to the longitudinal axis of the probe.

4.   The probe according to Claim 3, wherein said second magnetic field source comprises a radiation transmitting coil mounted such that it faces said angular sector, the geometrical center of the transmitting coil coinciding with the axis of magnetization of the magnets.

5.   The probe according to Claim 2, wherein said receiver comprises a radiation receiving coil mounted in said gap between the permanent magnet assemblies.

6.   The probe according to Claim 1, wherein the intensity of the static magnetic field in said high-gradient region monotonously decreases by about 50% within a distance from the probe edge of a dimension similar to a transverse dimension of the probe.

7.   The probe according to Claim 3, wherein said second magnetic field source is accommodated with respect to the first magnetic field source such that said sub-region is in the form of an angular sector of said imaging slice.

8.   The probe according to Claim 1, wherein the frequency bandwidth of the simultaneously excited nuclei spins in the sub-region is about 5%-200%, and typically 10%, of the mean value of the magnetic resonance frequency for the nuclei in the sub-region.

9.   The probe according to Claim 1, wherein said receiver is operable with a frequency bandwidth matching the operating frequency bandwidth of said second time-varying magnetic field.

10.   The probe according to Claim 9, wherein said frequency bandwidth of the receiver is about 5%-200%, and typically 10%, of the mean value of the magnetic resonance frequency for the nuclei in the sub-region.

11.   The probe according to Claim 1, and also comprising a third magnetic field source of a kind capable of creating a third, angular coordinate varying magnetic field within said at least sub-region of the second time-varying magnetic field.

12.   The probe according to Claim 11, wherein said third magnetic field source includes a pair of parallel solenoidal coils, which are arranged such that their axes

are parallel to the longitudinal axis of the probe, and located inside the probe close to the probe edge.

13.   The probe according to Claim 12, wherein the parallel coils are wound in an opposed manner.

14.   The probe according to Claim 11, wherein said parallel coils are wound on ferrite cores, thereby increasing the intensity of the produced magnetic field.

15.   The probe according to Claim 11, wherein said angular coordinate varying field is characterized by a monotonous change of the field intensity with the angular coordinate.

16.   The probe according to Claim 11, wherein said angular coordinate varying magnetic field is characterized by a substantially linear dependence of the field intensity on the angle within an imaging sector of a slice surrounding the probe in a plane perpendicular to a longitudinal axis of the probe.

17.   An MRI system for imaging a region of interest in a medium, wherein the image of the region of interest along at least one dimension of said region of interest is created by utilizing a means for producing at least a first magnetic field gradient along said at least one dimension, the system comprising the probe of Claim 1.

18.   An intravascular catheter for use in an MRI system, the intravascular catheter comprising the probe of Claim 1, and a position stabilizing assembly for fixing a relative position of a distal end of the catheter relative to surrounding blood vessel walls during an image acquisition process.

19.   An imaging method for creating an image of a region of interest in a medium, the method comprising the steps of:

(i)   applying a first, static magnetic field to the region of interest, said static magnetic field having a profile defining a high gradient magnetic field region such that a relative change of the static magnetic field intensity, $\Delta B_0/B_0$, in this region is of about 1% and more that the mean value of the static magnetic field intensity B_0;

(ii)   applying a second, time-varying magnetic field to at least a sub-region of said static magnetic field region, the time-varying magnetic field having a wide frequency range to thereby simultaneously excite nuclei spins of various resonance frequencies in said sub-region to generate NMR signals characterized by an NMR frequency bandwidth of 1% and more of a mean frequency value of said bandwidth;

(iii)   receiving the NMR signals and generating data indicative thereof.

20.   The method according to Claim 19, wherein the frequency bandwidth of the simultaneously excited nuclear spins in the sub-region is about 5%-200%, and typically 10%, of the mean value of the magnetic resonance frequency for the nuclei in the sub-region.

21.   The method according to Claim 19, wherein at least one hundred NMR signals are acquired in one excitation series having duration of a typical transverse relaxation time (T_2).

22.   The method according to Claim 19, wherein the applying of said first static magnetic field results in the creation of the high-gradient magnetic field region in the form of an imaging slice of the region of interest.

23.   The method according to Claim 22, wherein said applying of the second time-varying magnetic field results in the excitation of said sub-region in the form of a sector of the slice-like region of the static magnetic field.

24.   The method according to Claim 23, wherein the excitation of the nuclei in said sector by said time-varying magnetic field comprising the step of:

-    generating short transmission pulse bursts of sufficient power, thereby exciting successive sub-regions of said sector, each of the successive sub-regions including nuclei corresponding to a narrower resonance frequency band, as compared to that of the entire sector, the carrier frequency of each transmission pulse burst being selected according to a specific resonance frequency of the nuclei residing in the corresponding sub-region of the sector.

25. The method according to Claim 19, and also comprising the step of applying a third magnetic field to the medium in said static magnetic field region, to thereby create an additional imaging dimension of the region of interest.

26. The method according to Claim 25, wherein the third magnetic field is an angular coordinate varying magnetic field, the application of the third magnetic field resulting in phase encoding of nuclei spins within said sub-region of the time-varying magnetic field in an angular dimension.

27. The method according to Claim 25, wherein the third magnetic field is an angular coordinate varying magnetic field, the application of the third magnetic field resulting in narrowing the sub-region of the time-varying magnetic field in an angular direction.

28. The method according to any one of Claims 19-27, and also comprising the steps of applying said first static magnetic field so as to create successive high-gradient static magnetic field regions located in a common plane; and exciting nuclei within successive sub-regions of the high-gradient static magnetic field regions, thereby scanning and imaging the entire region of interest in the form of a circular slice.

29. The method according to any one of Claims 19-28, and also comprising the step of applying said first static magnetic field so as to create successive high-gradient static magnetic field regions located in different planes aligned along a common axis; and exciting nuclei within successive sub-regions of the high-gradient static magnetic field regions.
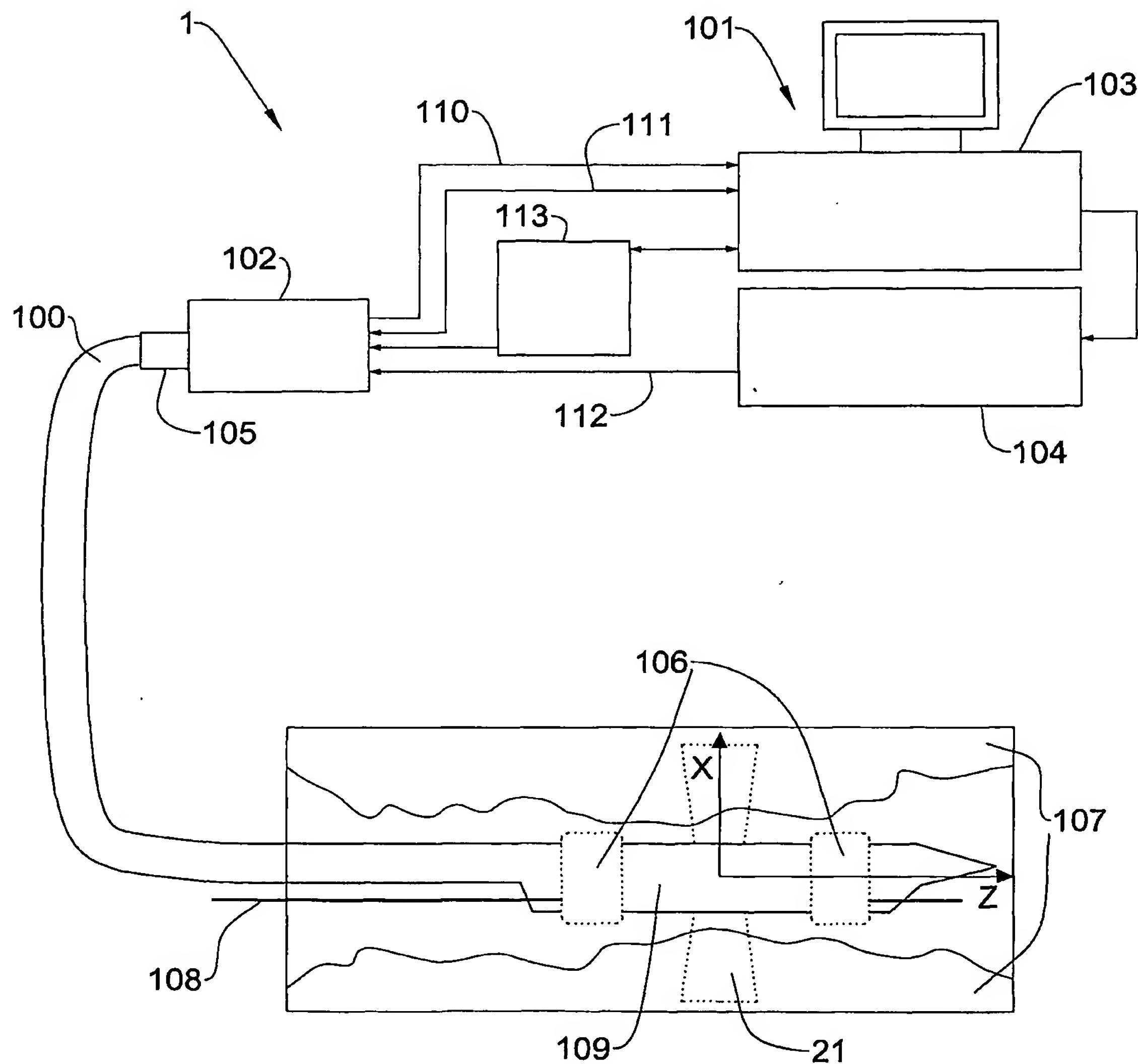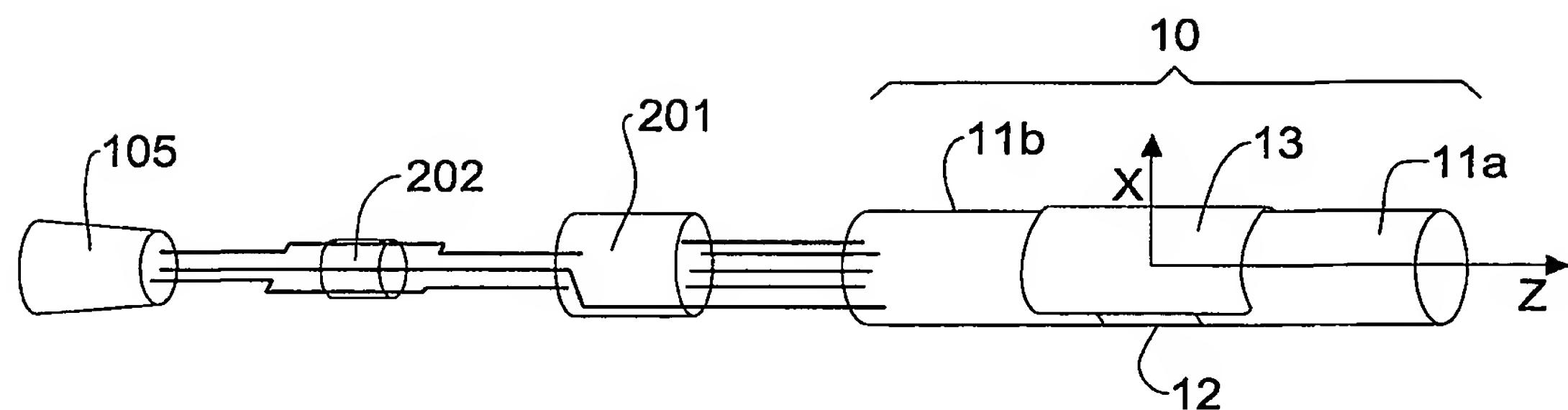
FIG. 1

FIG. 2
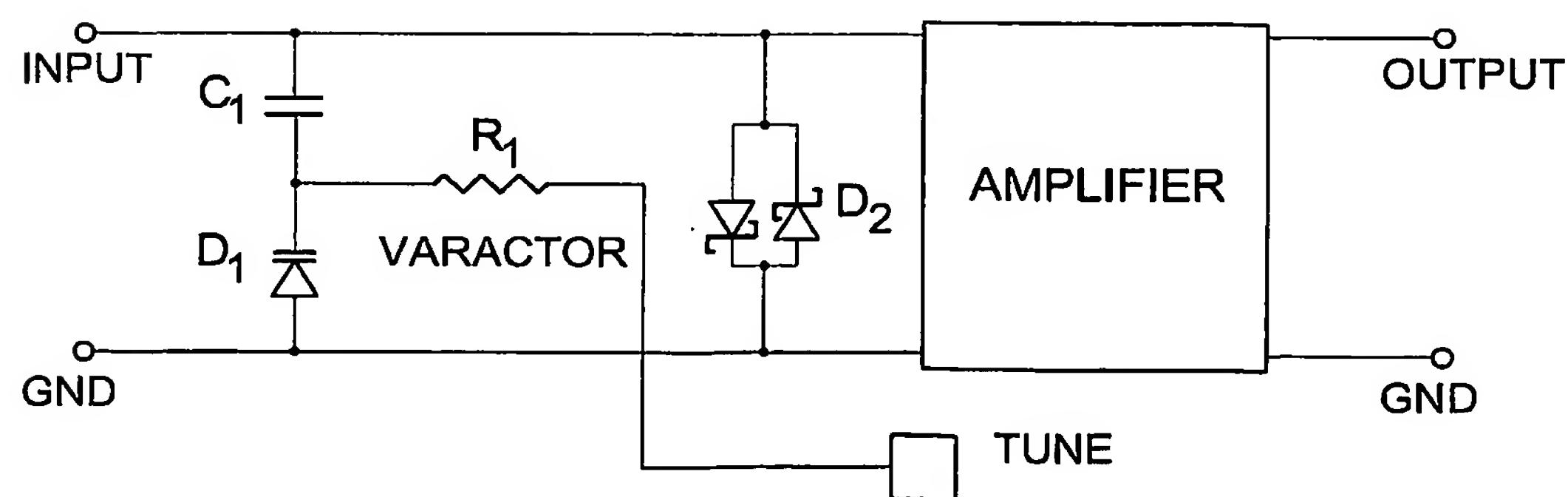
FIG. 3
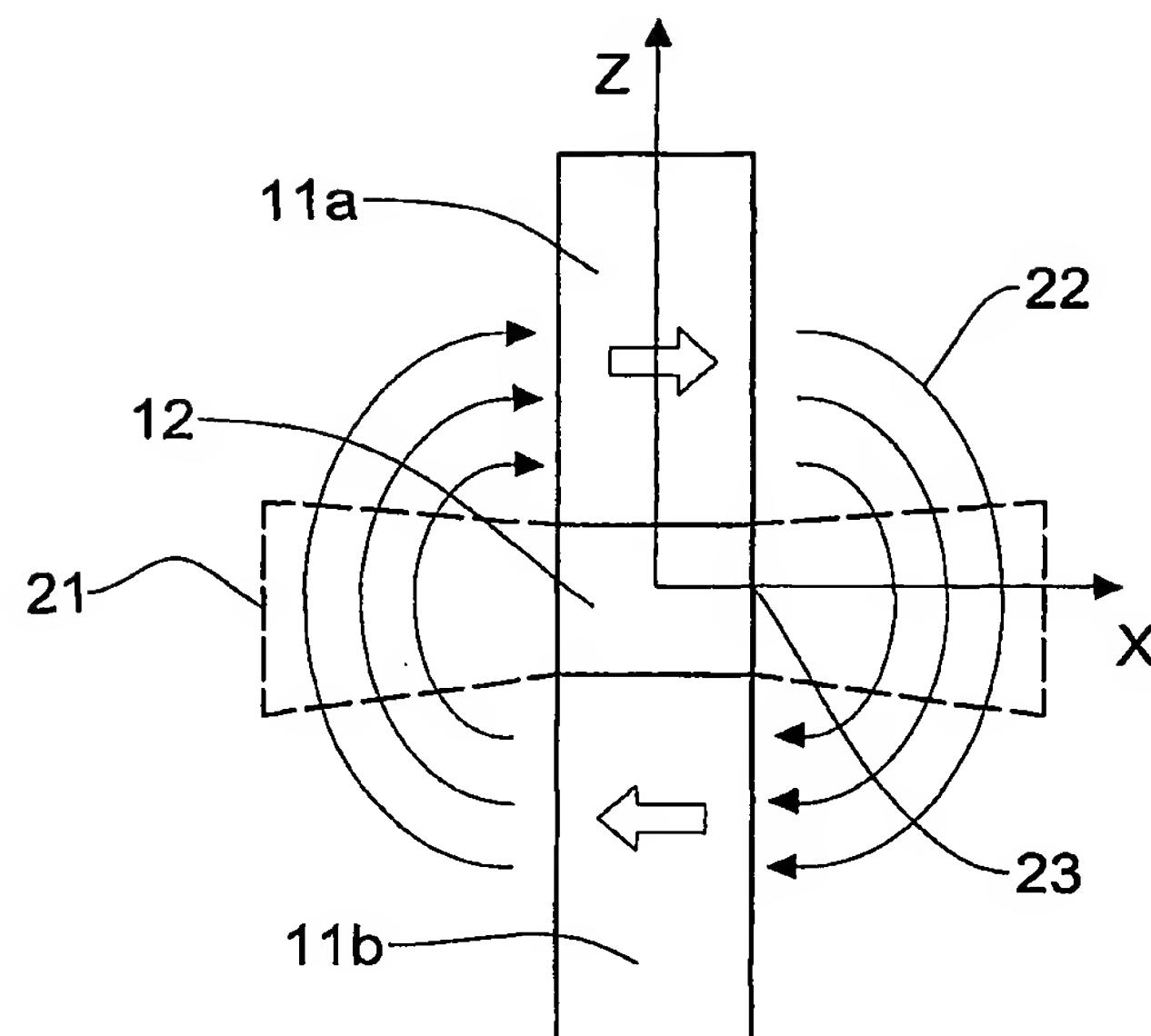
FIG. 4
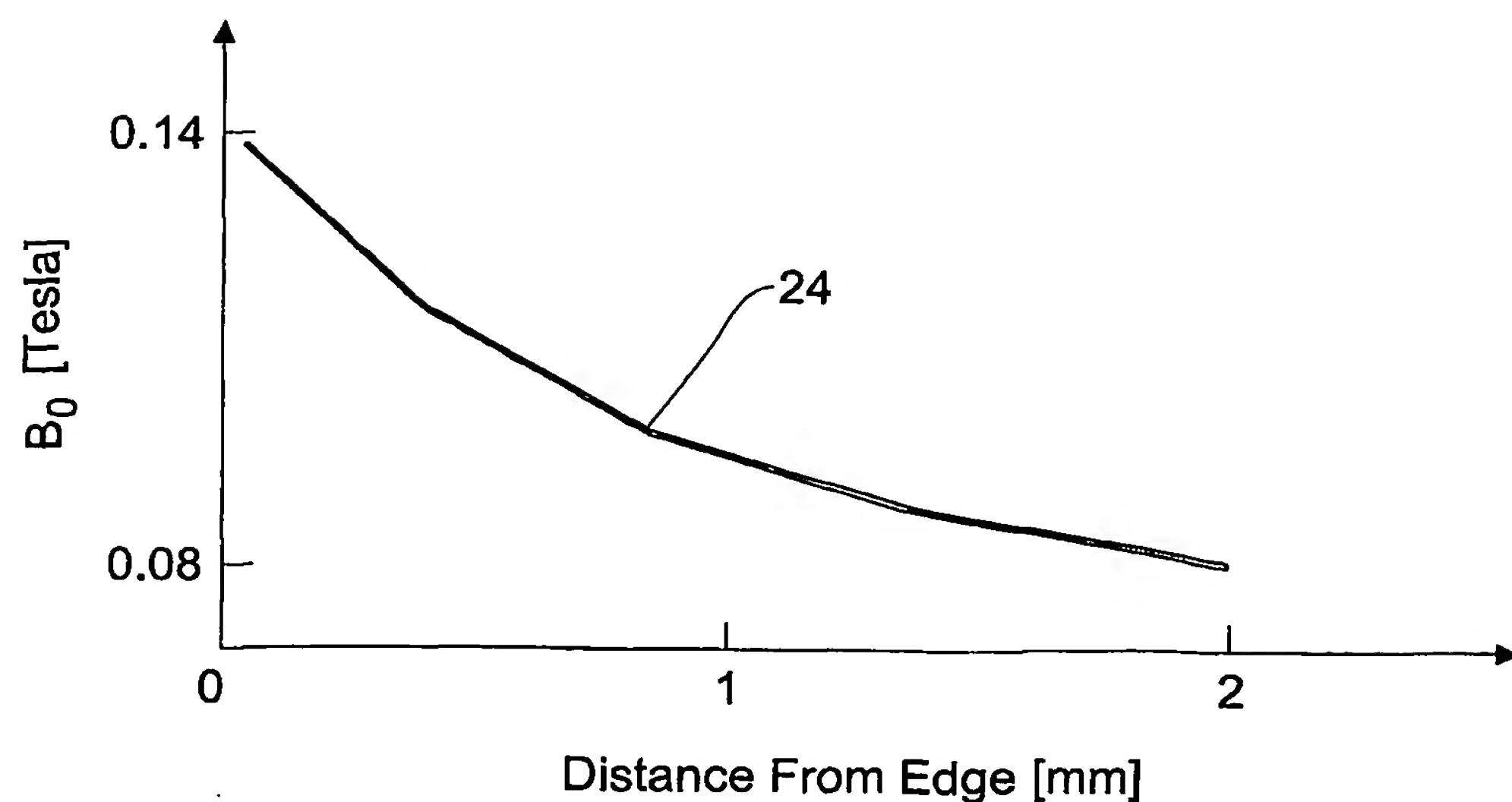
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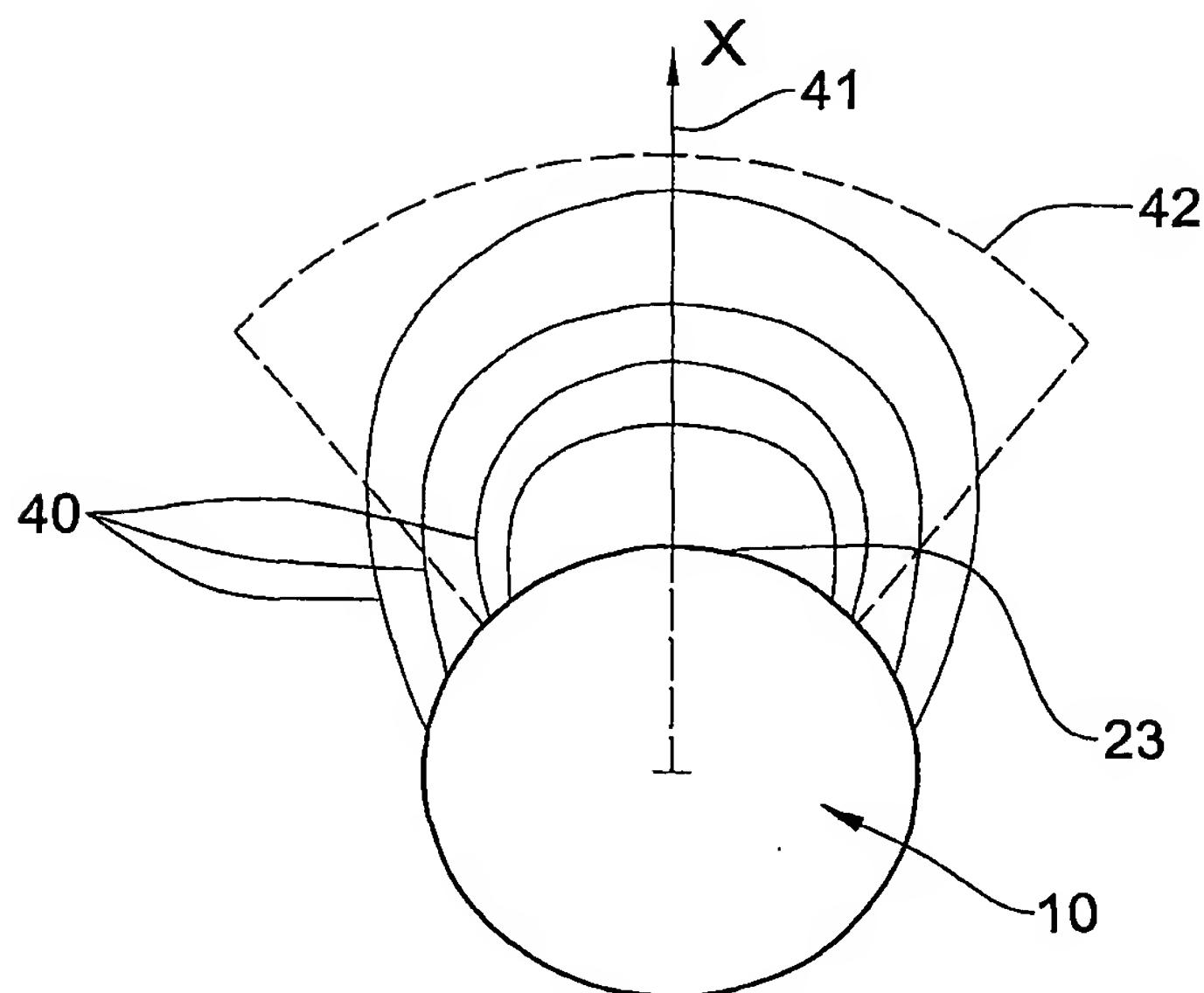
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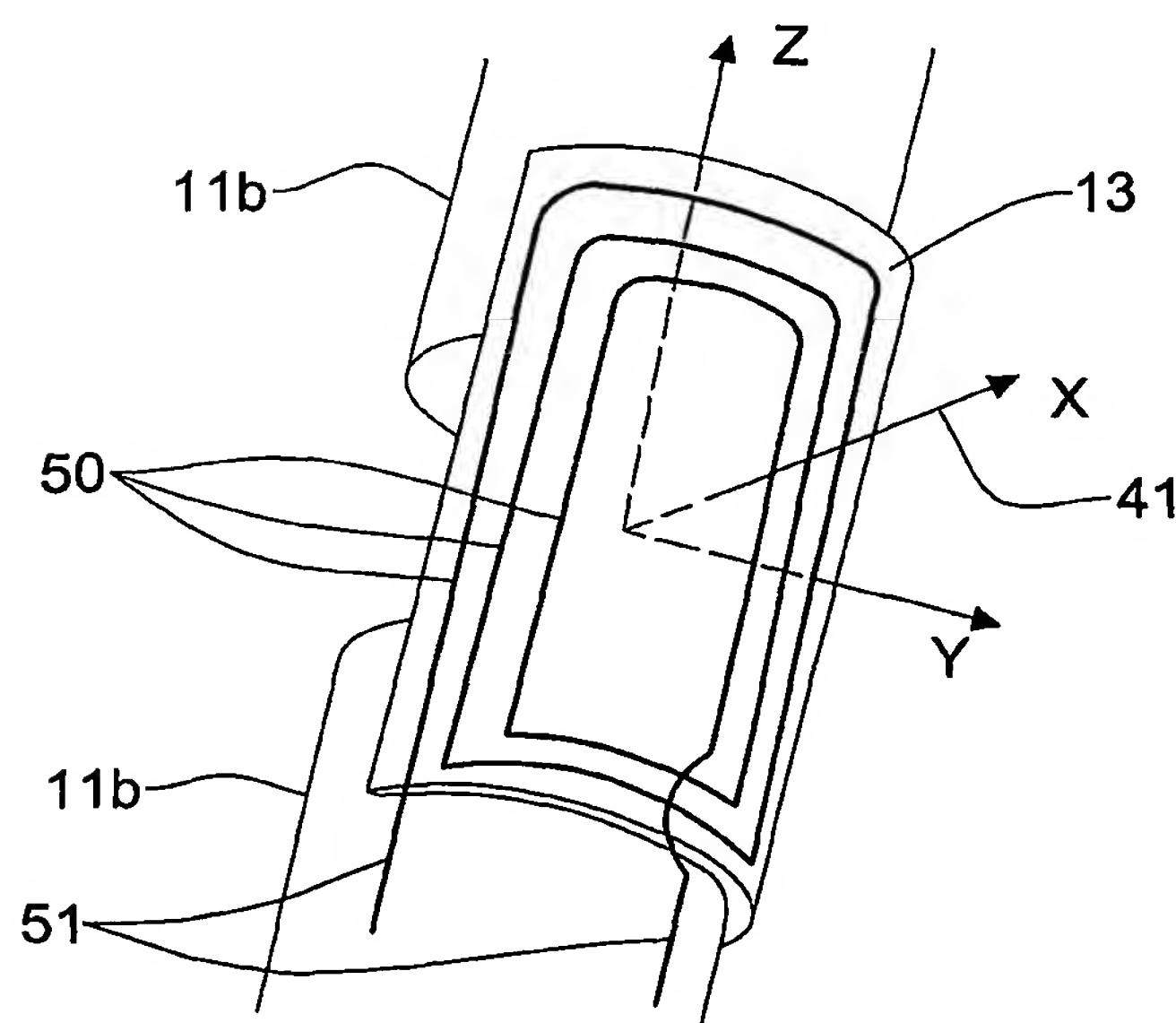
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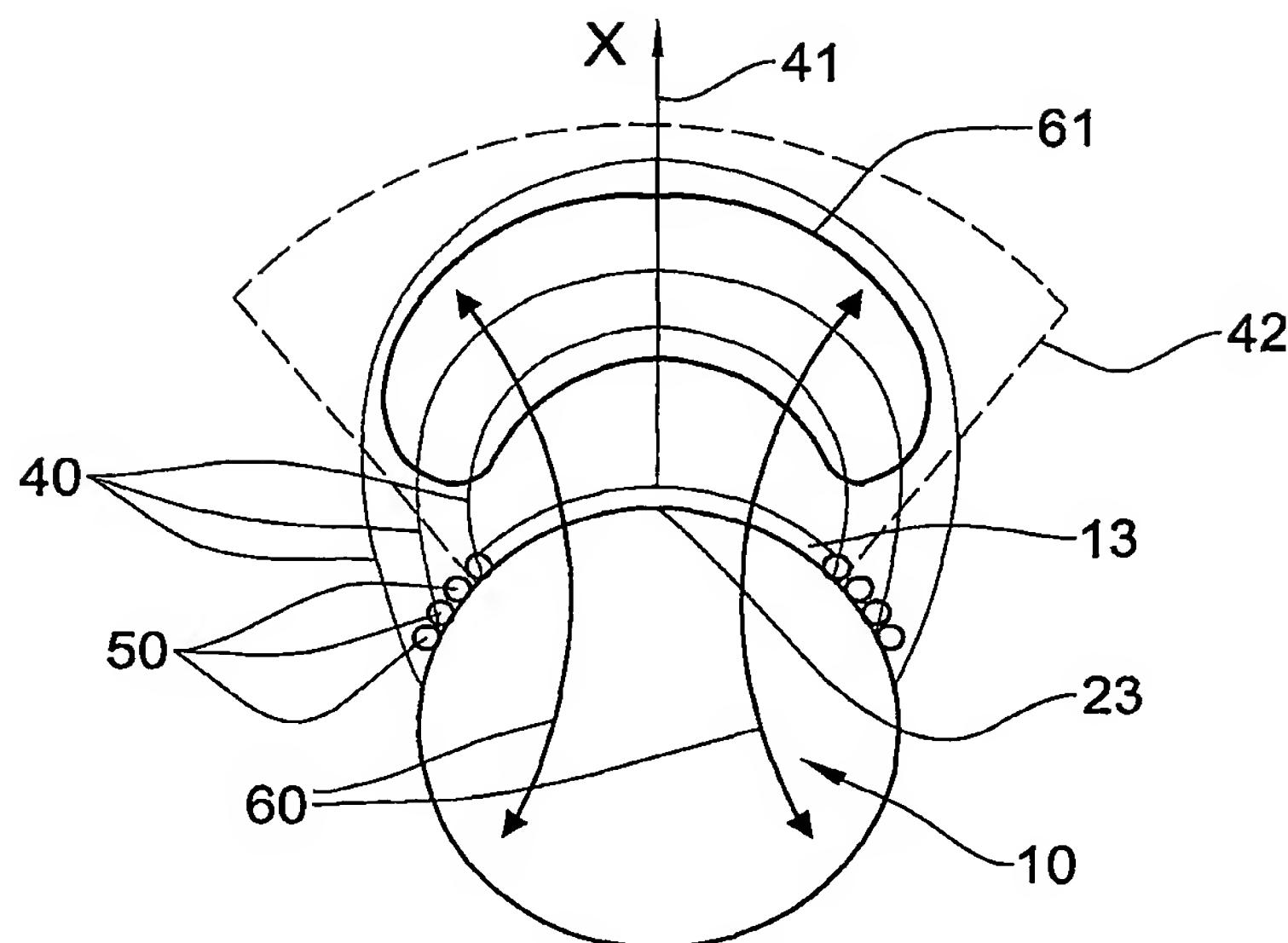
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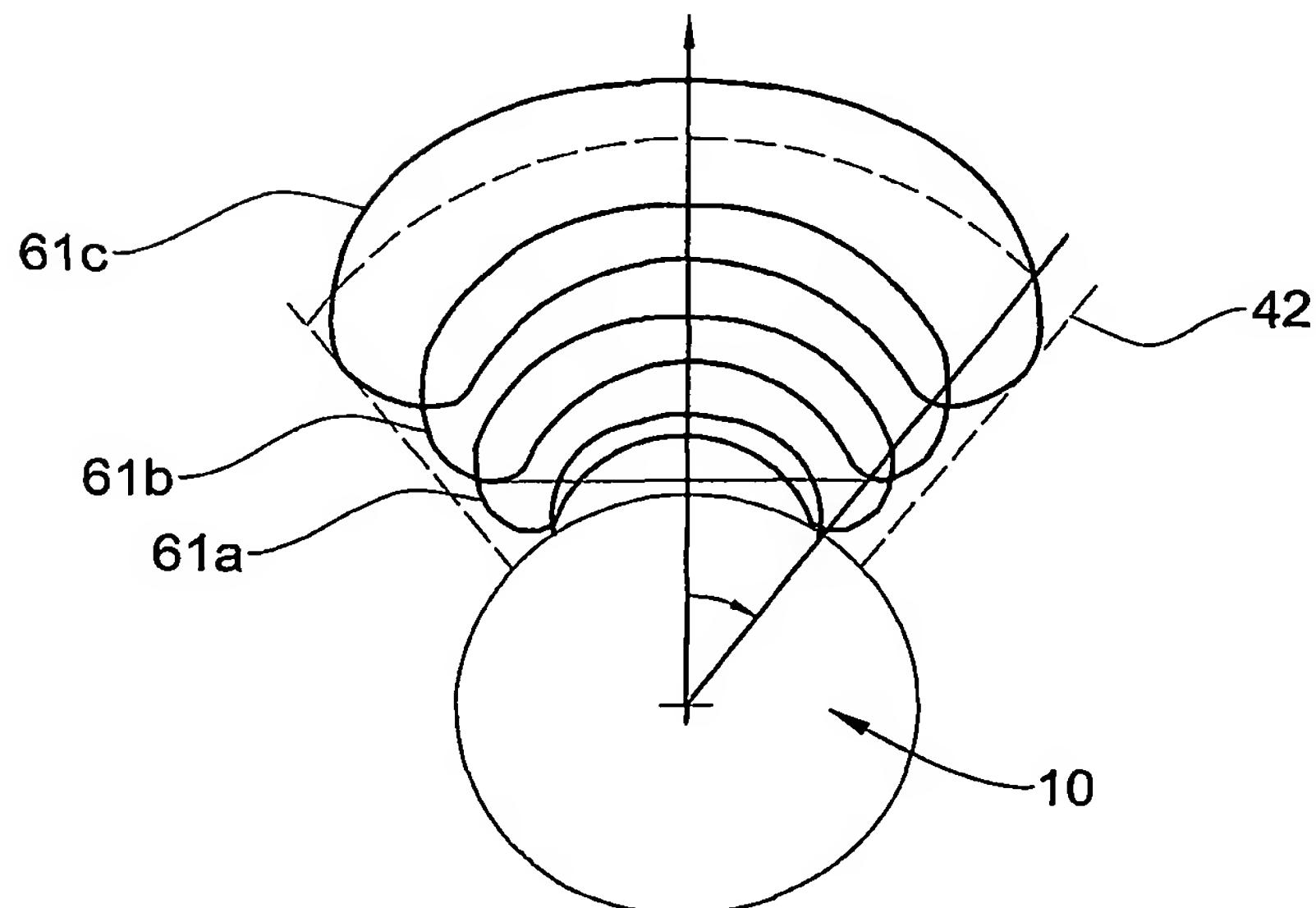
FIG. 9
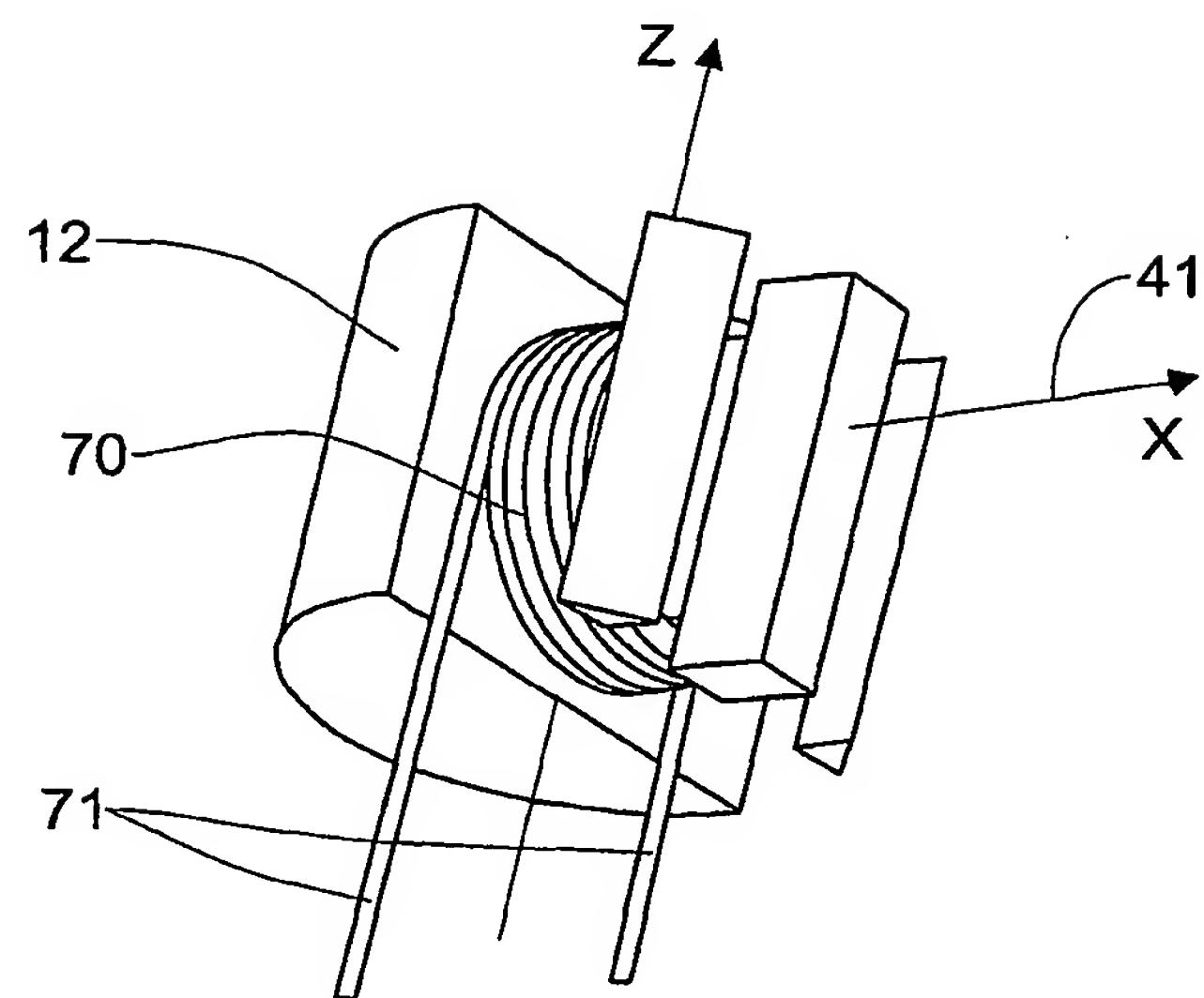
FIG. 10
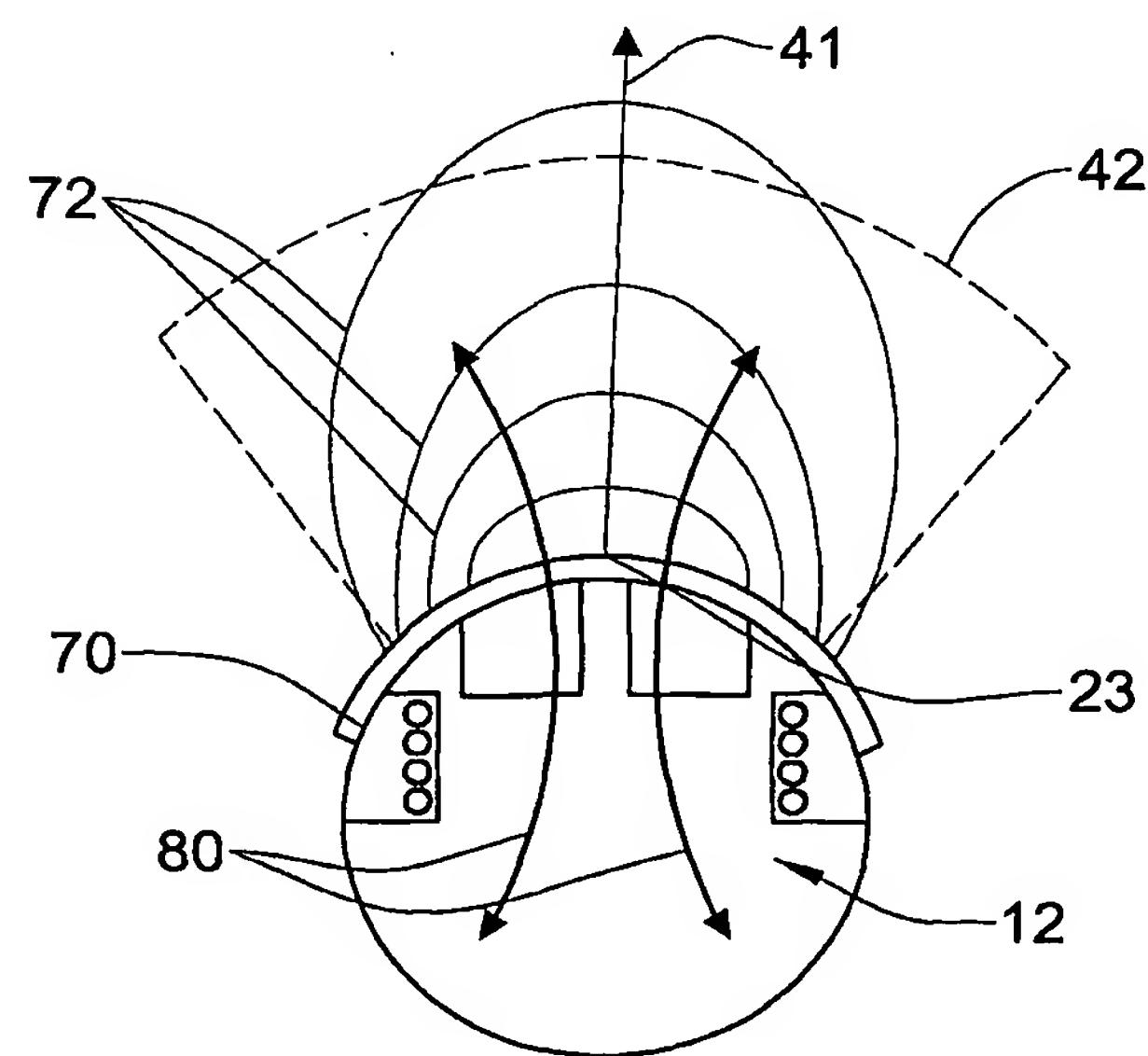
FIG. 11
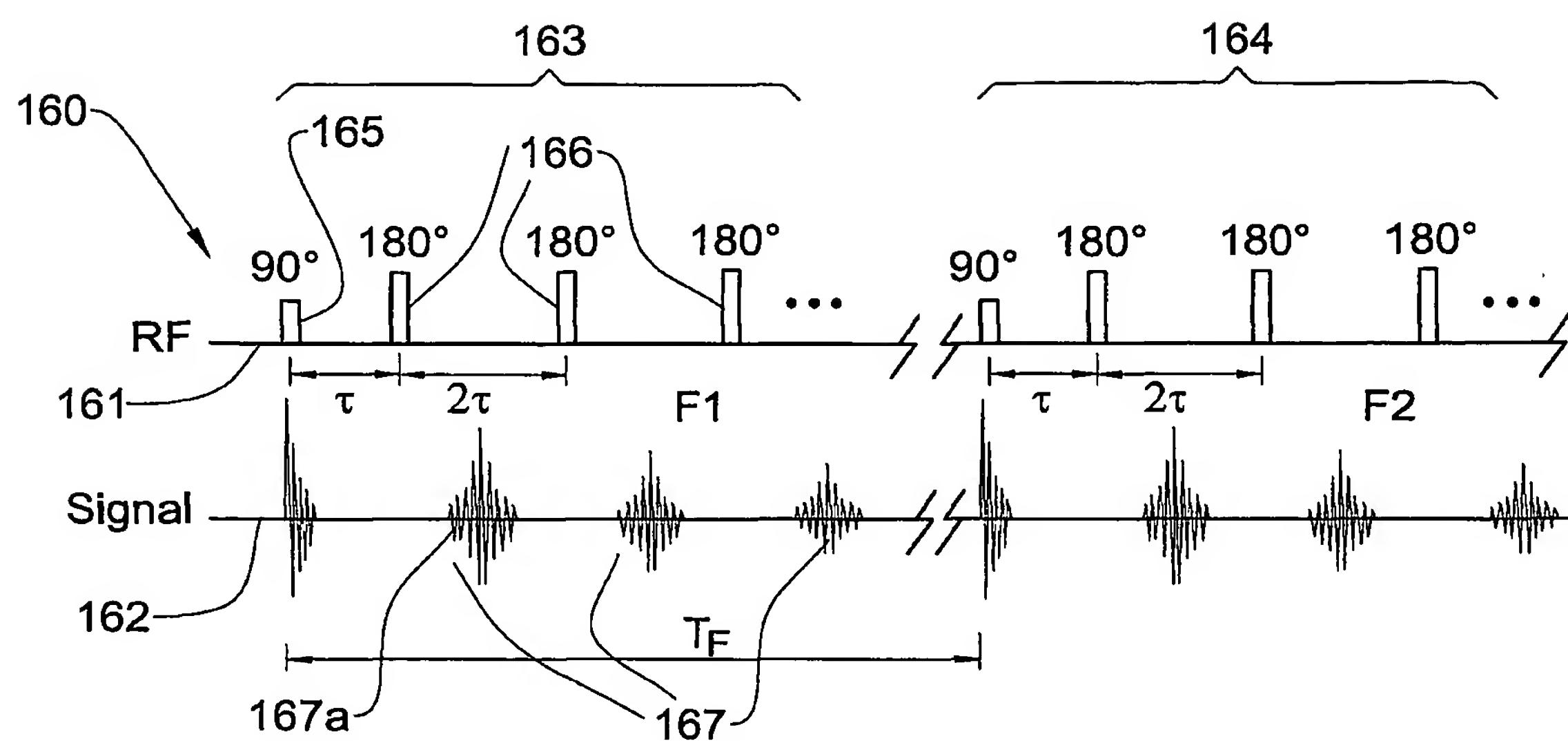
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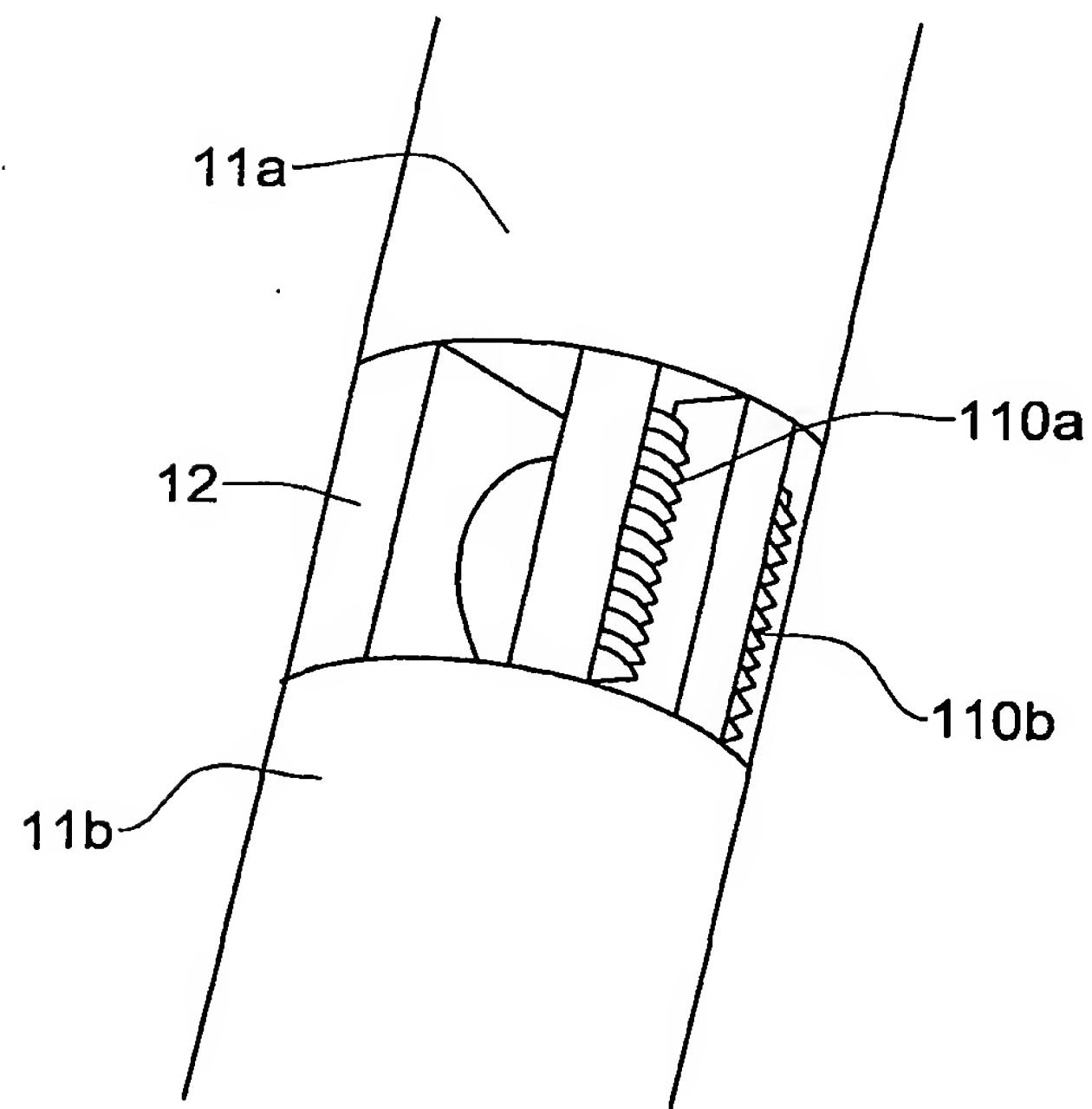
FIG. 13

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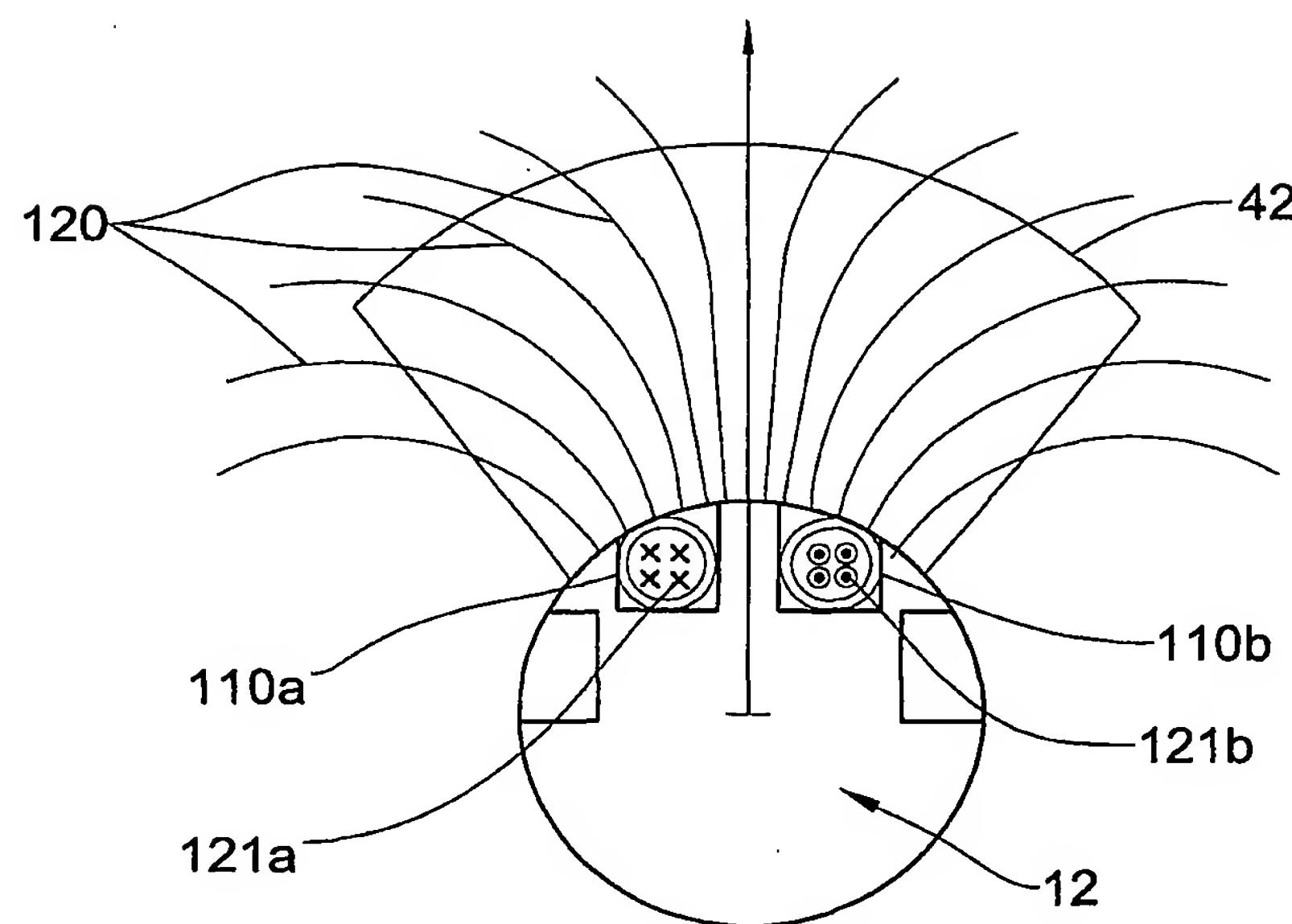
FIG. 14
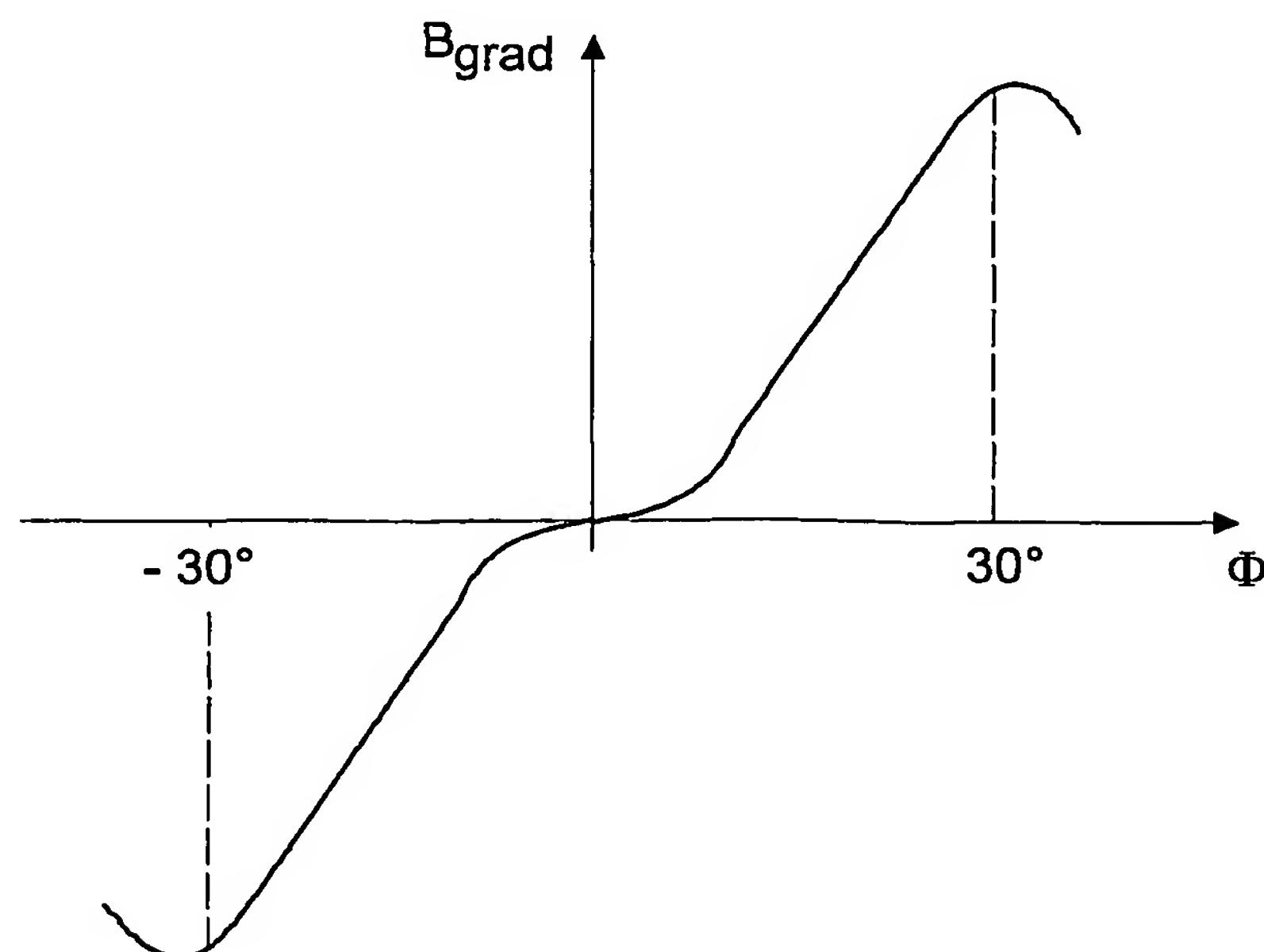
FIG. 15
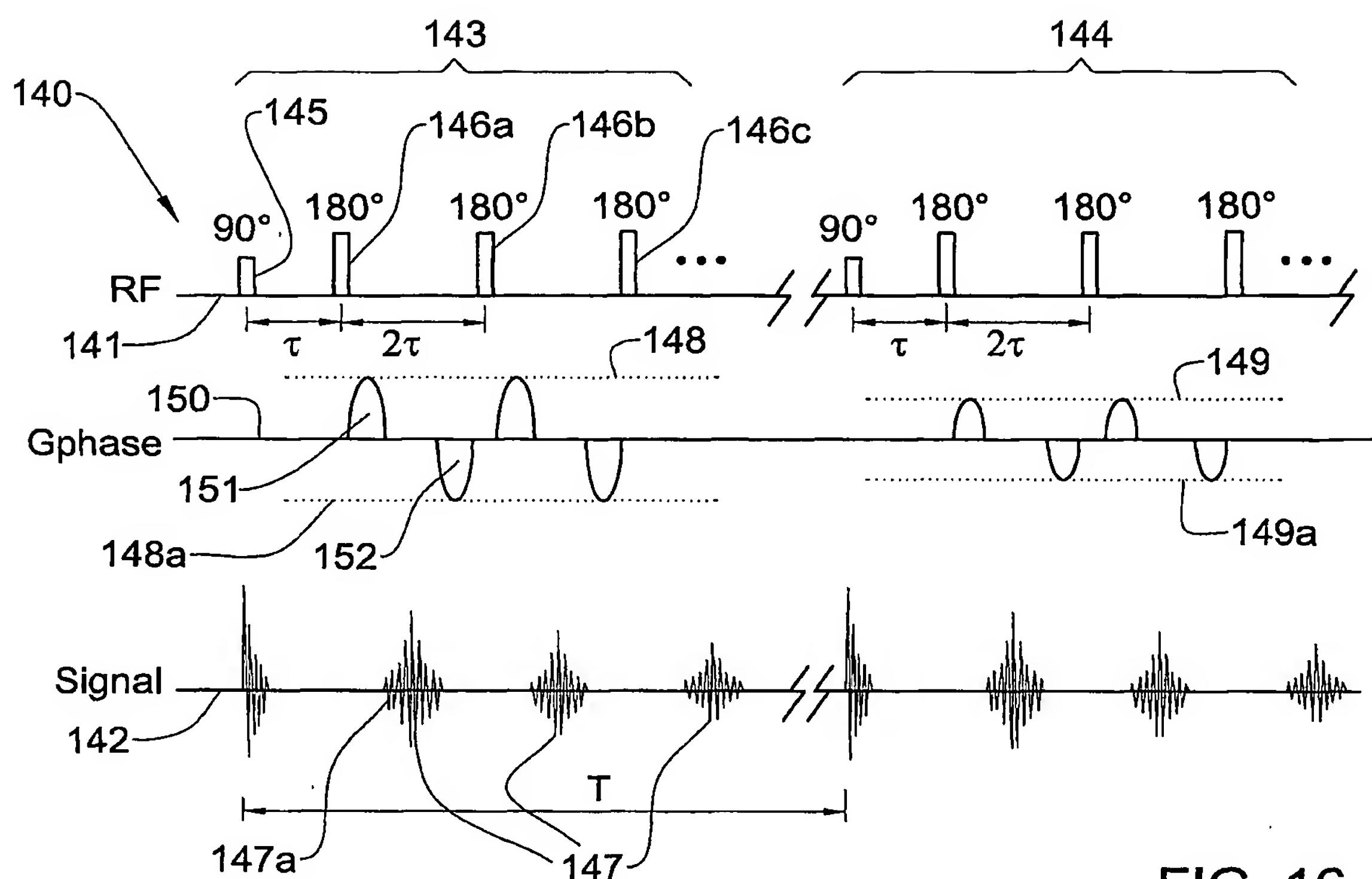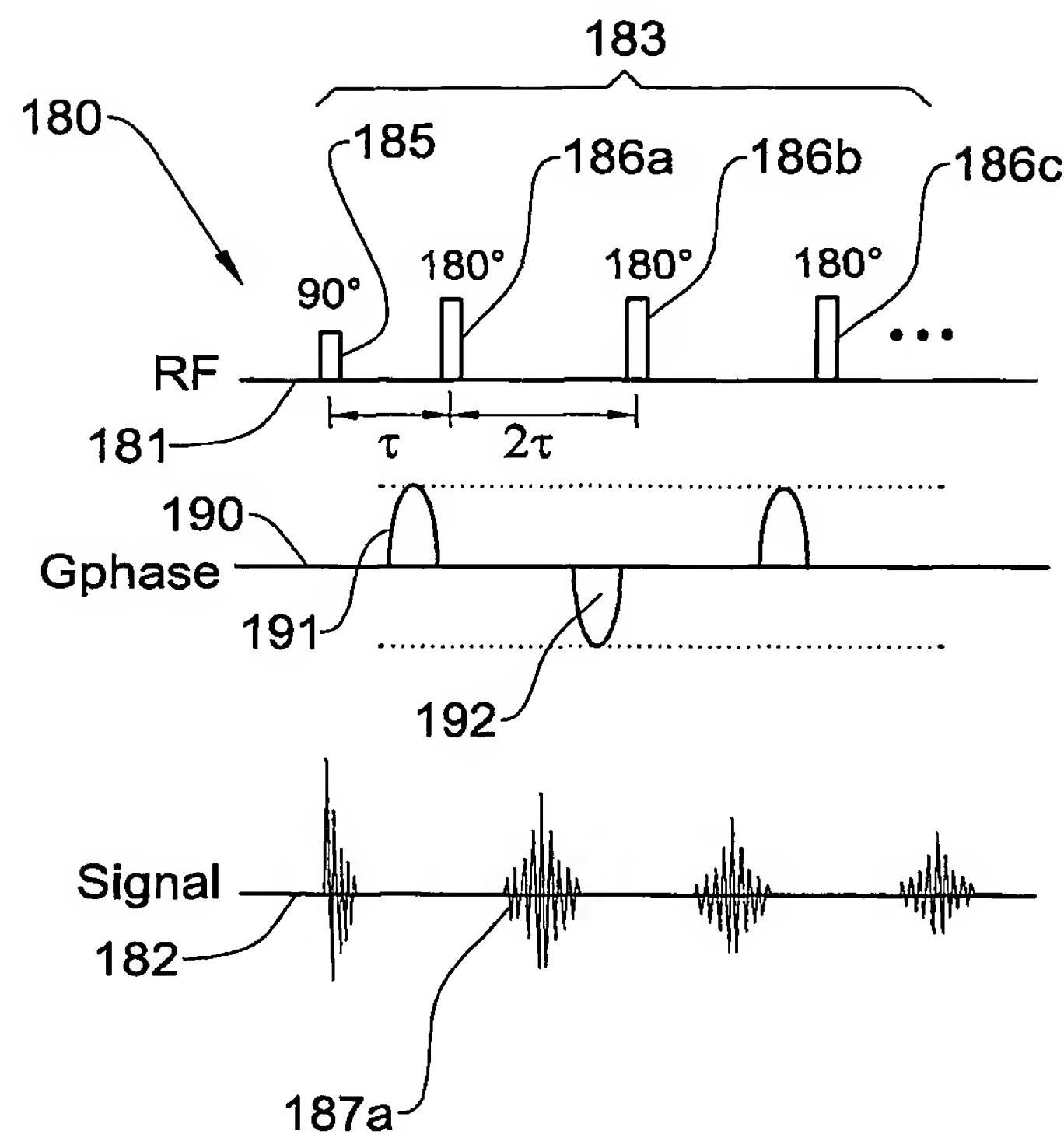
FIG. 16

FIG. 17

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FIG. 18

INTERNATIONAL SEARCH REPORT

A. CLASSIFICATION OF SUBJECT MATTER
IPC 7   G01R33/28   G01R33/44   G01V3/32   G01R33/383

According to International Patent Classification (IPC) or to both national classification and IPC

B. FIELDS SEARCHED

Minimum documentation searched (classification system followed by classification symbols)

IPC 7   G01R   G01V   G01N

Documentation searched other than minimum documentation to the extent that such documents are included in the fields searched

Electronic data base consulted during the international search (name of data base and, where practical, search terms used)

INSPEC, WPI Data, PAJ, EPO-Internal, MEDLINE

C. DOCUMENTS CONSIDERED TO BE RELEVANT

Category °	Citation of document, with indication, where appropriate, of the relevant passages	Relevant to claim No.
X	KLEINBERG R L ET AL: "NOVEL NMR APPARATUS FOR INVESTIGATING AN EXTERNAL SAMPLE" JOURNAL OF MAGNETIC RESONANCE, ACADEMIC PRESS, ORLANDO, FL, US, vol. 97, no. 3, 1 May 1992 (1992-05-01), pages 466-485, XP000273747 ISSN: 1090-7807 figures 1-6 ---	1,6, 8-10, 19-23
X	US 5 572 132 A (PULYER YULY M ET AL) 5 November 1996 (1996-11-05) cited in the application column 1, line 13 - line 17 column 5, line 7 -column 6, line 66 column 8, line 11 - line 34 ---	1,6, 8-19, 25-27
	-/--	

[X] Further documents are listed in the continuation of box C.   [X] Patent family members are listed in annex.

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Date of the actual completion of the international search	Date of mailing of the international search report
26 February 2002	05/03/2002

Name and mailing address of the ISA	Authorized officer
European Patent Office, P.B. 5818 Patentlaan 2 NL – 2280 HV Rijswijk Tel. (+31–70) 340–2040, Tx. 31 651 epo nl, Fax: (+31–70) 340–3016	Skalla, J

C.(Continuation) DOCUMENTS CONSIDERED TO BE RELEVANT		
Category *	Citation of document, with indication,where appropriate, of the relevant passages	Relevant to claim No.
X	US 4 714 881 A (GIVENS WYATT W) 22 December 1987 (1987-12-22) column 4, line 17 -column 5, line 42 ---	1-3,5-7
P,X	WO 01 42807 A (GOLAN EREZ ;TOPSPIN MEDICAL ISRAEL LTD (IL)) 14 June 2001 (2001-06-14) * see the whole document * ---	1-20
A	US 6 133 734 A (MCKEON DONALD C) 17 October 2000 (2000-10-17) column 5, line 1 -column 6, line 13 column 6, line 61 -column 7, line 23 -----	1,2

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			AU	1727101 A	18-06-2001
			WO	0142807 A2	14-06-2001
			US	2001011889 A1	09-08-2001
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